MONTGOMERY, MELISSA MEE HAE, Ph.D. The Effect of Lean Body Mass and Strength on Lower Extremity Energy Absorption Strategies during Landing. (2011) Directed by Dr. Sandra J. Shultz. 189 pp.

Lower amounts of lean body mass and strength in females compared to males have been proposed to result in greater relative task difficulty for females when asked to perform a standardized task and may explain sex differences in energy absorption strategies during landing maneuvers. The primary objective of this study was to determine the effect of lower extremity lean mass and eccentric muscle strength on lower extremity energy absorption strategies during a drop jump landing task. The secondary objective was to compare sex differences in energy absorption strategies when the task demands were equalized relative to the amount of lean mass available to dissipate kinetic energy upon landing. This was accomplished by separating out the effects of body composition and strength from other sex confounding variables by examining lower extremity lean mass and eccentric thigh strength in males and females matched by similar BMI's.

Thirty-five males were matched to 35 females on body mass index, and then completed body composition testing via dual energy x-ray absorptiometry, maximal eccentric strength testing via isokinetic dynamometry, and biomechanical assessment during a drop jump landing task. Each matched pair performed the drop jump landing task from two different heights: one at a standard height, and one at a height that equalized the relative task demands for the males relative to the females' standard height. The overall hypothesis was that less lower extremity lean mass relative to body mass would predict less lower extremity energy absorption during the deceleration of landing;

this relationship would be mediated by maximal eccentric thigh strength. Additionally, sex differences in energy absorption strategies were expected to diminish once the relative task demands were equalized.

The results showed that males had 42% more lower extremity lean mass ($p<0.001$) and produced 22% larger eccentric quadriceps ($p<0.001$) and 25% larger eccentric hamstring ($p<0.001$) peak torques than females. Analysis of the relationships between lean mass, eccentric strength, and energy absorption revealed significant positive relationships in females only, but these relationships did not increase with an increase in task difficulty. When comparing males and females on energy absorption strategies, males absorbed 44% more energy at the hip ($p=0.002$) than females before equalizing the task difficulty. After equalizing the task difficulty, the differences became larger with males absorbing 59%, 16%, and 22.5% more energy than females at the hip ($p<0.001$), knee ($p=0.038$), and ankle ($p=0.041$), respectively.

These results indicate that sex differences in energy absorption are not explained by sex differences in eccentric strength or relative task difficulty. More work is needed to determine additional factors that influence energy absorption strategies and which may further explain the sex differences in energy absorption strategies and the ultimate risk of injury.

THE EFFECT OF LEAN BODY MASS AND STRENGTH ON LOWER EXTREMITY ENERGY ABSORPTION STRATEGIES DURING LANDING

by

Melissa Mee Hae Montgomery

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Approved By

_Sandra J. Shultz________________________
Committee Chair

To my Family,

The opportunities you have provided have enabled me to achieve this accomplishment.

Thank you for your love, support and encouragement throughout all of my endeavors.

APPROVAL PAGE

This dissertation has been approved by the following committee of the Faculty of the Graduate School at The University of North Carolina at Greensboro.

Committee Chair ______________________
Sandra J. Shultz

Committee Members ______________________
Randy J. Schmitz

____
Laurie Wideman

____
Robert Henson

____
David H. Perrin

____
Date of Acceptance

____
Date of Final Oral Examination

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CHAPTER I

INTRODUCTION

Sport is the largest industry in the world (Broughton, 2008) and is a means of maintaining a lifestyle of health and well-being. In 2007, an estimated $14.7 billion (or 6.2% of the total dollars spent on sport) was spent for medical treatment of athletes alone (Broughton, 2008). Among these injuries, anterior cruciate ligament (ACL) injury remains one of the most costly injuries to treat, typically requiring reconstructive surgery and a lengthy rehabilitation. Although athletes are usually able to return to activity, many suffer lasting effects of this injury, particularly early onset osteoarthritis (Daniel et al., 1994; Lohmander, Ostenberg, Englund, & Roos, 2004; von Porat, Roos, & Roos, 2004), which can result in long term dysfunction and disability (Daniel, et al., 1994; Bjordal, Arnly, Hannestad, & Strand, 1997; Myklebust, Holm, Maehlum, Engebretsen, & Bahr, 2003; Lohmander, et al., 2004). Given the significant financial and health concerns associated with ACL injury, extensive prevention efforts have been undertaken. Yet despite these prevention efforts over the past decade, females remain 3-4 times more likely to sustain a non-contact ACL injury than their male counterparts (Arendt & Dick, 1995; Arendt, Agel, & Dick, 1999; Agel, Arendt, & Bershadsky, 2005; Mihata, Beutler, & Boden, 2006; Hootman, Dick, & Agel, 2007).

Experts agree that the cause for the sex disparity in injury rates is most likely due to sex differences in neuromechanical strategies (Griffin et al., 2006; Renstrom et al., 2008; Shultz, Schmitz, & Nguyen, 2008; Shultz et al., 2010). ACL injury most frequently occurs when trying to decelerate or change the momentum of the body (Myklebust, Maehlum, Engebretsen, Strand, & Solheim, 1997; Myklebust, Maehlum, Holm, & Bahr, 1998; Boden, Dean, Feagin, & Garrett, 2000; Olsen, Myklebust, Engebretsen, & Bahr, 2004; Krosshaug et al., 2007). During these types of activities, females typically utilize a characteristic "stiff" (Devita & Skelly, 1992; Lephart, Ferris, Riemann, Myers, & Fu, 2002; Decker, Torry, Wyland, Sterett, & Richard Steadman, 2003; Schmitz, Kulas, Perrin, Riemann, & Shultz, 2007) landing strategy which is often characterized by smaller hip and knee flexion angles with larger ground reaction forces (Huston, Vibert, Ashton-Miller, & Wojtys, 2001; Lephart, et al., 2002; Decker, et al., 2003; Salci, Kentel, Heycan, Akin, & Korkusuz, 2004; Yu, Lin, & Garrett, 2006; Chappell, Creighton, Giuliani, Yu, & Garrett, 2007; Schmitz, et al., 2007; Shultz, Nguyen, Leonard, & Schmitz, 2009). Stiff landings are also accompanied by greater quadriceps activation amplitudes (Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001; Chappell, et al., 2007), peak knee extensor moments (Chappell, Yu, Kirkendall, & Garrett, 2002; Salci, et al., 2004) anterior shear forces, and larger vertical ground reaction forces (vGRF) (Devita & Skelly, 1992), all of which are thought to place excessive strain on the ACL. Additionally, these stiff landings are associated with a reduced ability of the lower extremity muscles to absorb ground reaction forces during deceleration type maneuvers (Devita & Skelly, 1992; Zhang, Bates, & Dufek, 2000) and a shift to greater relative demands on the knee and ankle joints (versus the hip) to absorb these forces (Zhang, et al., 2000; Decker, et al., 2003; Schmitz & Shultz, 2010), both of which may expose passive structures to higher forces.

Although these sex-specific neuromechanical strategies have been extensively investigated and are well-described, we still have an incomplete understanding of what specific innate sex-related risk factors are influencing these strategies. Among the intrinsic anatomical and hormonal risk factors which have been proposed, one that remains relatively unexplored is the clear sex difference observed in body composition. This is despite the fact that males and females enter puberty with similar body size (Rogol, 2003; Wells, 2007), body composition (Rogol, 2003; Wells, 2007), and neuromechanical strategies (Hewett, Myer, & Ford, 2004; Quatman, Ford, Myer, & Hewett, 2006; Schmitz, Shultz, & Nguyen, 2009; Ford, Shapiro, Myer, Van Den Bogert, & Hewett, 2010) yet emerge substantially different in each, likely a result of hormonal changes that occur during maturation. Specifically, males produce more testosterone, which leads to gains in body mass primarily via gains in fat-free mass while females are exposed to greater estrogen levels leading to greater gains in fat mass (St-Onge & Bjorntorp, 2005; Wells, 2007; Loomba-Albrecht & Styne, 2009). These sex-dependent changes in body composition ultimately result in adult females possessing, on average, approximately 150- 200% more fat-mass than males and only 50-70% of their fat-free mass (Malina, 2005; Loomba-Albrecht & Styne, 2009) for a given total body weight. Concomitant with these body composition changes is the observation that boys (but not girls) experience an increase in power, strength, and neuromuscular control (Hewett, et al., 2004; Quatman, et al., 2006; Schmitz, et al., 2009). The concurrent timing of these events lends support to the notion that body composition is a plausible underlying factor that may drive the observed sex differences in neuromechanical strategies.

To date, limited work has examined the relationship between body composition and the ability to dissipate deceleration forces. This is important, as females with an above average body mass index (BMI; an estimate of body composition (Smalley, Knerr, Kendrick, Colliver, & Owen, 1990)) have been reported to have a 3.5 times greater risk of sustaining an ACL injury compared to those with an average BMI (Uhorchak et al., 2003). Because muscle mass is highly correlated with muscle strength (Fukunaga et al., 2001), the mechanism by which body composition may influence landing strategies in females is likely through decreased relative muscle strength as a result of reduced available fat free mass relative to total body mass. Consistent with having less relative total lean mass, and more specifically, less lower extremity lean mass (LELM), females also produce lower maximum quadriceps and hamstring torques (Lephart, et al., 2002; Salci, et al., 2004; Shultz, Nguyen, Leonard, et al., 2009). While these reduced torque capabilities have been associated with stiffer single- (Lephart, et al., 2002) and double- (Salci, et al., 2004) leg landings characterized by less knee flexion excursion (Lephart, et al., 2002), larger peak knee extensor moments and vertical ground reaction forces than males (Salci, et al., 2004), the direct role of body composition differences (thus available lower lean body mass and strength capabilities) has yet to be adequately explored. However, recent studies have shown that artificially increasing body mass index and reducing strength relative to body mass induces dangerous landing strategies in males and females alike. Specifically, loading the trunk with 10% of body mass results in a more erect landing position and larger normalized peak knee extensor moments during a stop-jump task (C. N. Brown, Yu, Kirkendall, & Garrett, 2005), and larger quadriceps and gastrocnemius forces (Kulas, Hortobagyi, & Devita, 2010), ground reaction impulses, knee extensor angular impulses, and increased energy absorption only at the knee (Kulas, Zalewski, Hortobagyi, & DeVita, 2008) during a double-leg drop landing. Although these findings were a result of the acute placement of additional mass and there were no analyses by sex, they do provide initial insight into the consequence of possessing less strength relative to body mass, and their potential influence on landing strategies. Hence, it is plausible that sex differences in joint stiffening and energy absorption strategies may be reflective of a lessened ability to produce eccentric muscle torques, and thus energy absorption (Zhang, et al., 2000; Decker, et al., 2003), to perform controlled decelerations, the types of activity associated with ACL injury (Boden, et al., 2000; Olsen, et al., 2004; Krosshaug, et al., 2007).

Statement of the Problem

Sex differences in neuromechanical strategies are thought to be primarily responsible (Griffin, et al., 2006; Renstrom, et al., 2008; Shultz, et al., 2008; Shultz, Schmitz, Nguyen, Chaudhari, et al., 2010) for the 3-4 fold greater risk of ACL injury in females compared to males (Arendt & Dick, 1995; Arendt, et al., 1999; Agel, et al., 2005; Mihata, et al., 2006; Hootman, et al., 2007). Several intrinsic risk factors are known to differ between males and females and thought to underlie these differences in neuromechanical strategies; body composition represents one major sex difference that has largely been ignored to date. Specifically, males possess more lean mass and greater strength than their female counterparts (Maughan, Watson, & Weir, 1983; Anderson, Dome, Gautam, Awh, & Rennirt, 2001). Since a body which is composed of a greater proportion of available lean muscle mass will likely possess a greater capacity to safely decelerate the body and maintain dynamic joint stability, it appears that body composition may be a crucial factor in our understanding of sex-specific neuromechanical strategies and injury risk.

Objective and Hypotheses

The primary objective of this study was to determine the effect of lower extremity lean mass and eccentric muscle strength on lower extremity energy absorption strategies during a drop jump landing task. The secondary objective was to compare sex differences in energy absorption strategies when the task demands were equalized relative to the amount of lean mass available to dissipate kinetic energy upon landing. This was accomplished by separating out the effects of body composition and strength from other sex confounding variables by examining lower extremity lean mass and eccentric thigh strength in males and females matched by similar BMI's. These matched pairs then performed a drop jump landing task from two different heights: one at a standardized height, and one at a height that equalized the relative task demands for the males relative to the females' standardized height. The overall hypothesis was that less lower extremity lean mass relative to body mass would predict less lower extremity energy absorption during the deceleration of landing; this relationship would be mediated by maximal eccentric thigh strength. Additionally, sex differences in energy absorption strategies were expected to diminish once the relative task demands were equalized. Specifically, the following hypotheses were examined:

***Hypothesis 1a*:** Less lower extremity lean mass will predict less energy absorption.

Hypothesis 1b: The relationship between lower extremity lean mass and energy absorption is mediated by maximal eccentric muscle strength.

Hypothesis 2: The relationship between eccentric thigh torque and energy absorption will be stronger in females vs. males due to the relative greater task demands for females at both heights.

Hypothesis 3: Increasing the task demand will cause alterations in energy absorption strategies.

Hypothesis 3a: Compared to males, females will absorb more energy about the knee during the $\text{Height}_{\text{STD}}$ condition.

***Hypothesis 3b*:** Equalizing the relative task demands (by comparing $\text{Height}_{\text{STD}}$ in females vs. $\text{Height}_{\text{EQU}}$ in males) will result in similar energy absorption strategies between size-matched males and females.

***Hypothesis 3c*:** Increasing task demands will result in greater changes in energy absorption strategies in females compared to males from the $\text{Height}_{\text{STD}}$ to the $\text{Height}_{\text{EQU}}$ because of the greater overall task demands on females vs. males.

Assumptions and Limitations

1. Results from this dissertation can only be generalized to the highly trained and athletic population studied when performing a drop jump landing task.
2. Dual-energy X-Ray Absorptiometry (DXA) is a reliable and valid method for assessing lower extremity lean mass.
3. The Phase Space IMPULSE motion analysis system and Bertec force platforms are valid and reliable devices for kinematic and kinetic measurements, respectively.
4. The trunk and lower extremity can be represented by a series of rigid segments that rotate about a joint; the forces acting upon these joints can be estimated with inverse dynamics solutions.
5. The Law of Conservation of Energy holds true; that the potential energy which a body possesses at a height is equal to the kinetic energy which is generated upon a fall and that with which ground contact is made.
6. Energy absorption represents the eccentric work performed by the musculature during the deceleration phase of a landing.
7. The drop jump landing task is predominantly composed of stretch-shortening muscle actions, and has a negligible isometric component.
8. Participants will exert maximal effort during isokinetic strength testing and vertical jump component of the drop jump landing.

Delimitations

1. Participants will be limited to athletic males and females who regularly participate (≥3x/week) in athletic activities which include jumping, landing, and quick deceleration with change of direction.
2. Participants will be required to be healthy, and void of any history of lower extremity orthopedic surgery, injury to the knee ligaments or cartilage, current lower extremity injury or pain. Female participants must not be pregnant.
3. Following familiarization, participants must be able to successfully and consistently perform the drop jump landing task and the maximal strength testing in order to participate.
4. Only two heights will be studied in order to separate out the effects of sex, lower extremity lean mass, and strength on landing strategies.

Operational Definitions

Body Composition: the tissue components which comprise the body; divided into lean, fat, and bone tissues

Body Mass Index (BMI): ratio of body mass and height^2; calculated as body mass (kg) divided by body height^2 (m)

Dominant Limb: the self-selected stance leg when kicking a ball for maximum distance

Drop Jump Landing Task: a task which includes a double-leg vertical drop landing, followed immediately by a maximal vertical jump and subsequent double-leg landing

Foot contact: the frame when vertical ground reaction force reaches or exceeds 10N

Joint Energy Absorption (EA): the eccentric work (J) performed by the musculature of a particular joint; calculated as the integration of the negative joint power curve during the deceleration phase of landing using the following formulas:

- Power (J)= Moment (Nm) x Angular Velocity (rad/s)
- Moment (Nm): calculated from inverse dynamics: (Force (N) x Moment arm (m))
- Angular Velocity (rad*s^{-1}): 1st derivative of position at each time interval (Position x Time)

Kinetic Energy (KE): the amount of energy (J) generated through movement of a body; equal to potential energy

Landing Phase: the period of time ranging from initial foot contact to peak center of mass (COM) displacement

Lower Extremity Lean Mass (LELM): the amount of non-fat, non-bone tissue (kg) located in the lower extremity region, which includes the foot, shank, and leg (including the gluteo-femoral region) (see Appendix A for anatomical boundaries)

Moment: an angular force (Nm) which causes a rotation about an axis; calculated as the product of Force (N) and Moment Arm (m)

Potential Energy (PE): the amount of energy (J) that a body possesses by virtue of its location above the ground; calculated as the product of body mass (kg), gravitational acceleration (m/s^2), and height above the ground (m)

Region of Interest (ROI): method of identifying body segments from a DXA scan for the reason of quantifying the composition in only the specified area; segments are bound by standard anatomical landmarks (see Appendix A for definitions of individual regions)

Relative Joint Contribution: the percentage (%) of total energy absorption performed by each of the lower extremity joints of the dominant leg (hip, knee, and ankle); calculated by dividing the individual joint energy absorptions by the total energy absorption of the lower extremity

Relative Task Difficulty: represents the amount of lower extremity lean mass (kg) available relative to the amount of potential energy (J) when standing at a vertical distance from the ground

Torque: the amount of angular force (Nm) exerted by the musculature

Total Energy Absorption (EA_{TOT}): the sum of the eccentric work (J) performed by the musculature of the hip (EA_{HIP}), knee (EA_{KNEE}), and ankle (EA_{ANK}) of the dominant leg

CHAPTER II

REVIEW OF THE LITERATURE

Introduction

Sport is a means of maintaining a lifestyle of health and well-being and is also the largest industry in the world. In 2007, an estimated 238 billion dollars were spent on sport in the United States alone, of which approximately $14.7 billion (or 6.2% of the total dollars spent on sport) were used for medical treatment of athletic injuries (Broughton, 2008). Among these injuries, injuries to the anterior cruciate ligament (ACL) remain one of the most costly injuries to treat, typically requiring reconstructive surgery and a lengthy rehabilitation. Although athletes are typically able to return to activity following these treatments, they are still subject to early onset osteoarthritis (Daniel, et al., 1994; Lohmander, et al., 2004; von Porat, et al., 2004), which can result in long term dysfunction and disability (Daniel, et al., 1994; Bjordal, et al., 1997; Myklebust, et al., 2003; Lohmander, et al., 2004). Given the significant financial and health concerns associated with this injury, extensive prevention efforts have been undertaken. Yet despite these efforts, females remain 3-4 times more likely to suffer an ACL injury than their male counterparts (Arendt & Dick, 1995; Arendt, et al., 1999; Agel, et al., 2005; Mihata, et al., 2006; Hootman, et al., 2007).

While it appears that females demonstrate neuromechanical strategies which may be responsible for the sex difference in injury rate, we do not fully understand what causes females to adopt these strategies (Griffin, et al., 2006; Renstrom, et al., 2008; Shultz, et al., 2008; Shultz, Schmitz, Nguyen, Chaudhari, et al., 2010). This review will present what is currently known regarding the incidence and mechanism of ACL injury, as well as the proposed risk factors which may precipitate the sex bias in ACL injury risk. This knowledge will then be assembled to examine body composition as a potential underlying factor that may contribute to the sex differences in neuromechanical strategies that are thought to place females at greater risk for ACL injury.

Current Theories for Sex Differences in ACL Injury Rates

The sex bias in ACL injury rates is well established in epidemiological literature, with females being far more likely to sustain an ACL injury than males. As such, much work has been performed to identify the differences between males and females which may explain or contribute to these discrepancies in injury rates. This section will first describe the known sex differences in ACL injury rates, followed by the current state of knowledge regarding the proposed risk factors that may place females at greater risk for ACL injury.

Epidemiology: Sex Differences in ACL Injury Rates

Since 1989, detailed data from the National Collegiate Athletics Association (NCAA) Injury Surveillance System (ISS) have consistently shown that females sustain ACL injuries at a higher rate than males, particularly in the sports of soccer and basketball (Arendt & Dick, 1995; Arendt, et al., 1999; Agel, et al., 2005; Mihata, et al., 2006; Hootman, et al., 2007). Additionally, these injuries in females are more likely to occur during a non-contact situation as compared to males (Arendt & Dick, 1995; Hootman, et al., 2007). In the first 5-year reporting period (1989-1993), for every 1000 athlete exposures, female athletes had a significantly higher rate of ACL injury than their male counterparts in soccer (0.31 vs. 0.13) and basketball (0.26 vs. 0.07) (Arendt & Dick, 1995). This pattern continued through the next 4 years (1994-1998) (Arendt, et al., 1999) and was further corroborated by a follow-up study spanning 13 years (1990-2002) which showed that ACL injury rates have remained stable in basketball and soccer (Agel, et al., 2005). Over this time span, female soccer players (0.31) were 2.8 times more likely to suffer an ACL injury than males (0.11) while the rate in female basketball players (0.27) versus males (0.08) was 3.6 times higher. The sex differences in injury rate are even more pronounced when considering only non-contact ACL injuries, where female soccer and basketball players, respectively, are 3.3 and 4.6 times more likely to suffer a non-contact ACL injury than their male counterparts. The most recent studies which extend the NCAA data through 2004 continue to confirm these earlier reports (Mihata, et al., 2006; Hootman, et al., 2007).

Mechanisms of ACL injury

Mechanisms of ACL injury have been examined using self-recall, healthcare records (Arendt & Dick, 1995; Myklebust, et al., 1997; Myklebust, et al., 1998; Arendt, et al., 1999; Boden, et al., 2000; Agel, et al., 2005; Hootman, et al., 2007) and video documentation (Boden, et al., 2000; Olsen, et al., 2004; Krosshaug, et al., 2007).

Regardless of sex or sport, ACL injury primarily occurs via a non-contact mechanism (i.e. without direct contact to the knee) but often with indirect contact inducing a perturbation or awkward landing before the actual injury (Krosshaug, et al., 2007). At the time of injury, athletes are often performing rapid deceleration maneuvers preceding a landing or a change in direction (Myklebust, et al., 1998; Boden, et al., 2000; Olsen, et al., 2004; Krosshaug, et al., 2007), with the knee typically positioned near extension (Boden, et al., 2000; Olsen, et al., 2004) along with slight tibial rotation (Olsen, et al., 2004). Additionally, females may also demonstrate larger knee valgus angles around the time of injury (Krosshaug, et al., 2007).

Overall, these data show that ACL injuries typically occur via the same mechanism across sport and sex. The pressing question is why similar activities result in more ACL injuries in females compared to males. As such, much work has been performed to elucidate risk factors that may explain this sex bias in injury risk. Currently, the proposed risk factors are categorized at the macro-level as either being extrinsic or intrinsic in nature. Extrinsic factors relate to external factors (outside the body) such as weather and playing surface, while intrinsic factors are related to an individual's physical attributes such as hormonal influences and anatomical factors (Arendt, et al., 1999). The next section will focus on the proposed intrinsic risk factors which have been implicated in the sex bias in ACL injury.

Proposed Risk Factors for ACL Injury

While a number of extrinsic and intrinsic risk factors have been identified for ACL injury, sex differences in neuromechanical strategies during dynamic movement are often considered the most important determinant (Griffin, et al., 2006; Renstrom, et al., 2008; Shultz, et al., 2008; Shultz, Schmitz, Nguyen, Chaudhari, et al., 2010). Hence, sex differences in neuromechanical strategies have been investigated extensively and are well-identified. While other intrinsic risk factors, categorized as anatomical and hormonal in nature, likely underlie and drive these sex differences in neuromechanical strategies, the exact mechanisms by which this may occur are not clearly understood. The following sections will explore these potential interactions.

Sex Differences in Neuromechanical Strategies

ACL injury occurs when an external load places excessive strain on the ligament which causes mechanical failure. Because the ACL is the primary passive restraint to anterior tibial translation (Markolf, Mensch, & Amstutz, 1976), neuromechanical strategies which are dominated by quadriceps actions are thought to be especially risky since they may place strain on the ACL (Renstrom, Arms, Stanwyck, Johnson, & Pope, 1986; G. Li et al., 1999; DeMorat, Weinhold, Blackburn, Chudik, & Garrett, 2004). Previous work has shown that across different landing tasks, athletic (Devita & Skelly, 1992; Lephart, et al., 2002; Salci, et al., 2004) and recreationally-active females (Decker, et al., 2003; Chappell, et al., 2007; Schmitz, et al., 2007) utilize a characteristic landing strategy commonly described as a “stiff” landing (Devita & Skelly, 1992; Lephart, et al., 2002; Decker, et al., 2003; Schmitz, et al., 2007). These landings often feature an erect body posture (i.e. less hip and knee flexion at initial ground contact or in total joint excursions) with larger ground reaction forces (Huston, et al., 2001; Chappell, et al., 2002; Lephart, et al., 2002; Decker, et al., 2003; Salci, et al., 2004; Yu, et al., 2006; Schmitz, et al., 2007; Shultz, Nguyen, Leonard, et al., 2009) and have been observed during both double- and single-leg drop landings and stop-jump tasks. Additionally, these stiff landings are often accompanied by greater quadriceps activation amplitudes (Malinzak, et al., 2001; Chappell, et al., 2007; Shultz, Nguyen, Leonard, et al., 2009), peak knee extensor moments (Chappell, et al., 2002; Salci, et al., 2004; Yu, et al., 2006; Shultz, Nguyen, Leonard, et al., 2009), and anterior shear forces (Chappell, et al., 2002), all of which have been postulated to place strain on the ACL (Hewett, Stroupe, Nance, & Noyes, 1996; Hewett et al., 2005; Yu, et al., 2006). Hence, it is not surprising that females are often reported to demonstrate a "quadriceps dominant" pattern, characterized by lower hamstrings-to-quadriceps strength ratios (Hewett, et al., 1996; Myer et al., 2009), preferential recruitment of the quadriceps compared to the hamstrings during a sudden perturbation (Shultz et al., 2001), and greater quadriceps activation amplitudes during running (Malinzak, et al., 2001), cutting (Malinzak, et al., 2001; Hanson, Padua, Troy Blackburn, Prentice, & Hirth, 2008), and stop-jump tasks (Chappell, et al., 2007) compared to males.

The combined neuromechanical strategies described (i.e. shallow hip and knee flexion angles with larger contributions from the quadriceps) are thought to be a primary contributor to ACL injury risk based on extensive work with cadaveric models

(Renstrom, et al., 1986; Markolf et al., 1995; G. Li, et al., 1999; DeMorat, et al., 2004; Withrow, Huston, Wojtys, & Ashton-Miller, 2006). These studies have shown that when acting alone, contraction of the quadriceps introduces strain on the ACL (Renstrom, et al., 1986; G. Li, et al., 1999), particularly at shallow flexion angles of less than 30-45° (Renstrom, et al., 1986; G. Li, et al., 1999), and that an aggressive unopposed contraction may even be capable of causing injury (DeMorat, et al., 2004). However, with simultaneous hamstring contraction, strain on the ACL and anterior tibial translation (ATT) can be reduced if the knee is flexed to at least 60 degrees (Renstrom, et al., 1986; G. Li, et al., 1999). More recently, Withrow et al (Withrow, et al., 2006) expanded on these cadaver studies by introducing an impact force to the knee which simulated a drop landing, and examined knee biomechanics in response to controlled levels of quadriceps and hamstring forces. Their findings revealed that ACL strain was strongly related to the change in quadriceps force and knee flexion angle after the impact, but not the impact force itself. Together, these studies provide strong evidence that suggest that the quadriceps alone are capable of inducing strain on the ACL at shallow knee angles, and that the hamstrings can help reduce that strain, but only at angles greater than 30-60°. However, these findings are limited to cadaveric experiments which used controlled loads to simulate quadriceps and hamstring contractions throughout a range of motion. As such, the loads exerted on the ACL in vivo are not clear, but it appears that when dynamic motions are performed with the knee in an extended position, large unopposed quadriceps actions place may excessive strain on the ACL, thus increasing the likelihood of injury.

Because of the complex interaction between neuromuscular, kinematic and kinetic variables which comprise the stiff landing strategies often observed in females, examining energy absorption patterns has been used to further investigate these landing patterns in a more integrative manner. Energy absorption describes the negative work or eccentric action of the lower extremity muscles during the deceleration phase of a landing which takes into account joint position, moments, and the time over which this action occurs (Devita & Skelly, 1992; McNitt-Gray, 1993; Schmitz, et al., 2007; Zhang, Derrick, Evans, & Yu, 2008; Schmitz & Shultz, 2010). In the few studies which have compared energetic strategies in males and females (Decker, et al., 2003; Schmitz, et al., 2007; Schmitz & Shultz, 2010), females have been reported to perform landing tasks with similar (Schmitz & Shultz, 2010) or lower (Schmitz, et al., 2007) amounts of total muscular energy absorption during double- and single-leg landings, respectively. However, regardless of the type of landing, females tend to use the knee and ankle to a greater extent to absorb landing forces, compared to males who tend use their hip extensors to a larger degree (Decker, et al., 2003; Schmitz, et al., 2007; Schmitz & Shultz, 2010). These divergent energy absorption patterns are thought to be the result of females demonstrating a more erect landing position at initial contact with more total joint excursion which results in a greater reliance on the knee and ankle for energy absorption (Decker, et al., 2003). Whether this energetic pattern in females is potentially more injurious is unknown.

While sex differences in neuromechanical strategies and the theoretical consequences of such strategies have been well described, we still have an incomplete understanding of the underlying causes which drive the sex-specific use of these strategies. However it is widely agreed that the cause is likely multifactorial in nature (Arendt, et al., 1999). The following section will discuss anatomical and hormonal factors which have been proposed to influence the observed sex differences in neuromechanical strategies.

Potential Hormonal and Anatomical Factors Underlying Neuromechanical Strategies

Perhaps one of the most obvious differences between adult males and females is their circulating sex hormone concentrations, whereby females 1) experience large fluctuations in estrogen and progesterone over the course of the menstrual cycle and 2) have higher absolute levels of estrogen and lower absolute levels of testosterone compared to males. These sex difference in absolute and cyclic hormone concentrations are thought to drive the major sex differences in anatomy which emerge following puberty. The following section will address the possible mechanisms by which 1) the acute cyclic changes in hormone concentrations in females and 2) the absolute sex differences in sex hormone levels may affect anatomical factors thought to be related to altered neuromechanics and injury risk.

Effect of Cyclic Changes in Sex Hormone Concentrations

Sex hormones have been implicated as a risk factor for ACL injury because of evidence suggesting that the risk of sustaining an ACL injury may occur in a time-dependent fashion across the menstrual cycle, with a larger proportion of injuries reported to occur during the pre-ovulatory compared to the post-ovulatory phases

(Renstrom, et al., 2008; Shultz, et al., 2008). While the exact phase and hormone profile that represents the greatest risk of injury remain unclear, primarily due to methodological differences in determining cycle phase, these studies suggest that normal, physiological changes in hormone concentrations over the course of the menstrual cycle may impact the musculoskeletal system in a way that alters injury risk potential. However, the specific mechanism(s) by which acute changes in hormone concentrations may impact the musculoskeletal system to influence neuromechanical strategies is unknown.

Two areas that have received much attention are cyclic effects on ligament laxity and muscle strength. Because the human ACL expresses receptors for estrogen, progesterone, testosterone, and relaxin (Liu et al., 1996; Hamlet, Liu, Panossian, & Finerman, 1997; Dragoo, Lee, Benhaim, Finerman, & Hame, 2003; Lovering & Romani, 2005; Faryniarz, Bhargava, Lajam, Attia, & Hannafin, 2006) and estrogen and testosterone receptors are also present in skeletal muscle (Lemoine et al., 2003; Wiik et al., 2003; Sinha-Hikim, Taylor, Gonzalez-Cadavid, Zheng, & Bhasin, 2004), this suggests that sex hormones have the ability to interact with the structure and function of these tissues. The majority of work in this area did not find an effect of cycle phase on laxity (Karageanes, Blackburn, & Vangelos, 2000; Beynnon et al., 2005; Eiling, Bryant, Petersen, Murphy, & Hohmann, 2007). However, in the studies that have carefully determined cycle phase with serial hormone measurements, it has been shown that some females experienced cyclic changes in anterior knee laxity (Shultz, Kirk, Johnson, Sander, & Perrin, 2004; Shultz, Sander, Kirk, & Perrin, 2005; Shultz, Gansneder, Sander, Kirk, & Perrin, 2006). Among these studies, laxity was reported to be lowest shortly after menses when both estrogen and progesterone are at their nadirs and highest during the early luteal phase once estradiol levels peak and progesterone begins to rise (Shultz, et al., 2005), implicating the influence of estrogen on cyclic knee laxity. Because greater anterior knee laxity and general joint laxity have been identified as risk factors for future ACL injury (Uhorchak, et al., 2003), these later findings suggest that hormone-mediated increases in laxity may place females at an increased risk of injury compared to their male counterparts who do not experience cyclic hormone changes. While the mechanism for this relationship are not well defined at this point, recent studies (Park, Stefanyshyn, Ramage, Hart, & Ronsky, 2009b, 2009a) suggest that the changes in laxity observed across the menstrual cycle in some women may be sufficient to disrupt normal joint neuromechanics and potentially impair dynamic joint stability.

Because sufficient torque generation around the knee is crucial to maintaining joint stability during dynamic situations, changes in muscle strength due to cyclic changes in hormone levels have also been studied. As with joint laxity, these findings are equivocal with some investigators reporting cyclic changes in quadriceps and hamstring strength across the cycle (Sarwar, Niclos, & Rutherford, 1996; Bambaeichi, Reilly, Cable, & Giacomoni, 2004), while others have not (Lebrun, McKenzie, Prior, & Taunton, 1995; Gür, 1997; Janse de Jonge, Boot, Thom, Ruell, & Thompson, 2001; Fridén, Hirschberg, & Saartok, 2003; Montgomery & Shultz, 2010). The bodies of literature which have attempted to relate acute hormonal fluctuations to laxity and muscle strength is therefore filled with considerable disagreement as to the direction of these relationships. This is most likely due to methodological differences in menstrual cycle phase determination, which often fail to appreciate the considerable inter-individual differences in hormone magnitude and timing (Shultz, et al., 2004). This limitation in study designs has hindered our understanding of acute hormone fluctuations and cyclic changes in laxity and muscle strength, and further work is needed.

Hormonally-Induced Anatomical Differences between Males and Females

Another likely mechanism by which sex hormones impact knee joint neuromechanics is via anatomical differences in the lower extremity. These anatomical differences develop during maturation in response to the absolute differences in hormonal milieu under which males and females operate. In particular, at the time of puberty, males begin producing large amounts of testosterone, while the primary sex steroid in females is estrogen. These hormone changes are primarily responsible for the appearance of secondary sex characteristics, specifically the differences in body size and composition between males and females (Rogol, 2003; Wells, 2007).

During childhood, boys and girls are similar in size and body composition until the age of approximately 10-12 years (Malina, 2005). Shortly thereafter, both girls and boys undergo rapid increases in height and body mass during puberty (Wells, 2007). In regard to body mass, the increase in body mass in females that is due to fat-free mass levels off by age 15-16, with primary increases in body mass due to fat-mass thereafter (Rogol, 2003; Malina, 2005). The increase in adiposity, particularly in the gluteo-femoral region is thought to be largely due to the constant exposure to estrogen (Wells, 2007; Loomba-Albrecht & Styne, 2009). Conversely, males continue to gain body mass during this time primarily via gains in fat-free mass due to increases in testosterone (St-Onge &

Bjorntorp, 2005; Wells, 2007; Loomba-Albrecht & Styne, 2009) until age 19-20 (Rogol, 2003), while fat-mass levels off by 13-15 years old (Malina, 2005). Together, these hormonal changes result in the largest pubertal sex differences in body composition emerging around age 14-16, with females possessing approximately 150% more fat-mass than males and only 70% of the fat-free mass (Malina, 2005). The end result of these pubertal changes is that adult males and females possess similar body mass indices (BMI; mass*height^{-2}) (Wells, 2007), but females have approximately twice the body fat and half of the lean mass making up their total body mass compared to males (Loomba-Albrecht & Styne, 2009).

Concomitant with these pubertal changes in body size and composition is that boys (but not girls) experience an increase in power, strength, and neuromuscular control during maturation (Hewett, et al., 2004; Quatman, et al., 2006; Schmitz, et al., 2009). It is probably not by chance that these anatomical and neuromuscular changes emerge coincidentally with the pubertal changes in neuromechanics which have been observed in cross-sectional (Hewett, et al., 2004; Schmitz, et al., 2009) and longitudinal (Quatman, et al., 2006; Ford, et al., 2010) studies. Specifically, these studies show the divergence in neuromechanical strategies as maturation progresses, with females demonstrating increasingly larger knee valgus angles (Hewett, et al., 2004; Quatman, et al., 2006; Schmitz, et al., 2009) and a reduced ability to attenuate landing forces (Quatman, et al., 2006) during the late/post pubertal stages versus pre-puberty. Since these neuromechanical changes coincide with body composition changes, these findings suggest that sex differences in body composition may play a role in the sex differences in neuromechanical strategies that have been observed. However, to date, the role of body compositional changes in relation to knee joint neuromechanics has not been assessed.

Additional Anatomical Factors

Beyond the more obvious hormone-induced changes in growth and body composition, several additional anatomical differences between males and females emerge during maturation and may contribute to altered neuromechanics and risk of ACL injury. Specific to the ACL itself, females have smaller ACLs (length, cross-sectional area, and volume) than males (Anderson, et al., 2001; Chandrashekar, Slauterbeck, & Hashemi, 2005; Chandrashekar, Mansouri, Slauterbeck, & Hashemi, 2006) even after adjusting for body mass (Anderson, et al., 2001). This is an important anatomical difference, especially in light of the report that ACL-injured subjects have smaller ACL's than their non-injured matched controls (Chaudhari, Zelman, Flanigan, Kaeding, & Nagaraja, 2009). Additionally, studies consistently show that females have greater joint laxity than males. Specifically, females possess greater anterior knee laxity (Rozzi, Lephart, Gear, & Fu, 1999; Uhorchak, et al., 2003; Beynnon, et al., 2005; Shultz, et al., 2005; Shultz et al., 2007), varus-valgus and rotational knee laxity (Shultz, et al., 2007; Shultz & Schmitz, 2009), and general joint laxity (Beighton, Solomon, & Soskolne, 1973; Uhorchak, et al., 2003; Shultz, et al., 2007). Females and males also differ in their skeletal alignment, including increased hip anteversion, quadriceps and tibiofemoral angles and greater genu recurvatum in females compared to males (Nguyen & Shultz, 2007). Thus, while this review will focus on the role that body composition may play in determining sex-specific landing strategies, it is acknowledged that other sex-dependent anatomical factors may be operating as well.

Anatomical Influences on Neuromechanical Strategies

At this time, we do not have a complete understanding of how anatomical differences that emerge during maturation may influence sex-specific neuromechanical strategies. However, recent work has shown that greater anterior knee laxity is related to greater amounts of anterior tibial translation during a simulated transition from nonweight-bearing to weight-bearing (Shultz et al., 2006) and greater energetic demands at the knee with concurrent decreases in hamstring muscle activation in females, but not males, during drop jumping . Likewise, females with above average frontal and transverse plane knee laxity demonstrate more knee valgus upon landing, perform larger hip adduction and IR excursions, and are exposed to longer periods of hip adduction and knee valgus moments, all with larger sEMG muscle activation amplitudes (Shultz & Schmitz, 2009; Shultz, Schmitz, Nguyen, & Levine, 2010). Collectively, these studies show that joint laxity allows for greater multiplanar joint motion which may encourage an “at-risk” posture during dynamic motion, possibly leading to injury. However, as it is apparent that joint laxity does not explain the majority of sex differences in neuromechanical strategies, it appears that additional anatomical factors may aid in elucidating these sex differences. Despite the large differences in body composition between females and males, we know very little regarding the consequence of decreased amounts of lower extremity lean mass relative to total body mass on sex-specific neuromechanical strategies.

Summary: Hormonal and Anatomical Influences

Concomitant with emerging sex differences in estrogen and testosterone levels during maturation, anatomical dimorphism begins to emerge between males and females. Perhaps not coincidentally, sex differences in neuromechanical strategies also develop during this time. Together, this suggests that sex hormones influence sex-specific anatomical differences, which may in turn influence neuromechanical strategies. While the sex difference in ligament laxity is the most well-documented and has been shown to be related to dangerous neuromechanical strategies, less is known about the relationship between other anatomical differences and neuromechanics. In particular, despite the large differences in body composition and strength between males and females, very little is known regarding their influence on sex-specific neuromechanical strategies and resultant injury risk. Because a body which is composed of a greater proportion of available lean muscle mass (i.e. that in males) will likely possess a greater capacity to safely control motion and protect the joints, understanding the influence on body composition is important to our understanding of the underlying causes of the observed sex differences in neuromechanical strategies.

Section Summary:
Current Theories for Sex Differences in ACL Injury Rates

Females are at greater risk for sustaining an ACL injury than their male counterparts (Arendt & Dick, 1995; Arendt, et al., 1999; Agel, et al., 2005; Mihata, et al., 2006; Hootman, et al., 2007). Currently, the expert consensus is that sex differences in neuromechanical strategies are mostly responsible for the discrepancy in ACL injury rates (Griffin, et al., 2006; Renstrom, et al., 2008; Shultz, et al., 2008; Shultz, Schmitz, Nguyen, Chaudhari, et al., 2010). Although these differences between males and females are well described, the underlying intrinsic factors which may contribute to these sex-specific strategies remain an enigma. Females characteristically perform landing maneuvers in a stiff manner, with less hip and knee flexion, larger peak knee extensor moments and quadriceps activation amplitudes, and with larger energetic demands at the ankle and knee versus the hip, compared to their male counterparts (Huston & Wojtys, 1996; Malinzak, et al., 2001; Chappell, et al., 2002; Lephart, et al., 2002; Decker, et al., 2003; Schmitz, et al., 2007; Shultz, Nguyen, Leonard, et al., 2009; Schmitz & Shultz, 2010). These sex differences in landing strategies appear to develop during puberty (Hewett, et al., 2004; Quatman, et al., 2006; Schmitz, et al., 2009; Ford, et al., 2010) at the same time as the hormonally-induced dimorphism in body composition emerges, resulting in males possessing a larger proportion of lean mass to total body mass than females (Rogol, 2003; Malina, 2005; Loomba-Albrecht & Styne, 2009). During this time, males also experience substantial gains in strength and power, whereas females continue to gain body mass without the concomitant gains in strength (Hewett, et al., 2004; Quatman, et al., 2006; Schmitz, et al., 2009). Perhaps not coincidentally, sex differences in ACL injury rates also appear at this time (Shea, Pfeiffer, Wang, Curtin, & Apel, 2004).

All together, the literature in this area suggests that hormonally-driven anatomical factors, particularly body composition, may in part explain sex differences in neuromechanical strategies. Specifically, since the maturation process leaves adult females with more body fat and less lean mass, females may have more difficulty in safely decelerating and controlling the body's momentum during activities where ACL injury is known to occur. Considering these potential associations, the role of body composition in sex-specific neuromechanical strategies merits further investigation. To that end, the next section will discuss the potential influence of body composition on lower extremity neuromechanics and risk of injury.

Potential Relationships Among Body Composition, Lower Extremity Biomechanics, and Injury

While individual relationships between body composition, lower extremity biomechanics, and injury have been investigated, these associations have not been investigated in an integrative manner. This section will attempt to incorporate this body of literature to demonstrate the potential for a lower amount of lean mass relative to total body mass to influence lower extremity neuromechanics in a way that may place the knee at increased risk for injury. This section will first discuss what is known regarding the individual relationships between 1) body composition, strength and injury and 2) body composition, strength, and lower extremity biomechanics. These relationships will then be integrated to build a theoretical framework by which body composition and strength may influence lower extremity neuromechanics.

Relationship between Body Composition and Strength

The mechanism by which body composition (i.e. available lean muscle mass relative to total body mass) affects neuromechanics is likely through muscle strength because lean mass is highly correlated to the force-producing capability of the muscle (Knapik, Staab, & Harman, 1996; Bamman, Newcomer, Larson-Meyer, Weinsier, & Hunter, 2000). Given the role of neuromuscular control in maintaining joint stability, the muscle's ability to generate a torque around a joint and resist external forces experienced during dynamic activity is crucial. Although it is clear that males possess greater absolute strength and strength relative to body mass, it appears that these differences in strength are largely due to the fact that males simply possess more lean mass than females. For example, it has been shown that the sex difference in strength disappears when maximal strength is normalized to muscle cross-sectional area (Maughan, et al., 1983) or muscle volume (Akagi et al., 2009), but not body mass (Anderson, et al., 2001). When comparing 25 young males and females (mean age = 25 years), Maughan et al (Maughan, et al., 1983) showed that males produced higher peak quadriceps MVICs than females, even when normalized to total body mass. However, when peak torques were normalized to each participant's thigh cross-sectional area (measured by computed tomography), the sex differences in quadriceps MVIC disappeared, with both males and females demonstrating a significant correlation between quadriceps MVIC and muscle cross-sectional area (r=0.59 and 0.51, respectively). Likewise, Akagi et al (Akagi, et al., 2009) found no sex or age differences in elbow flexor MVIC when normalized to muscle volume measured by magnetic resonance imaging (men: r=0.76; women: r=0.93). These studies support a strong positive relationship between muscle size and strength. However, because the strength of muscle contraction is also influenced by neural mechanisms, we do not know whether these relationships are robust to contraction type as well. That is, there is no evidence to suggest or deny that the relationships between dynamic muscle strength and muscle size are the same as that which has been observed during an isometric contraction. Therefore, more work is needed to clarify the relationship between lean mass and a dynamic muscle contraction which more closely resembles that which is performed during sport-like activities where ACL injury typically occurs (e.g. eccentric strength). Establishing these relationships will in turn help us better understand the relationship between body composition, neuromechanics, and the resulting risk of injury.

Body Composition and Strength as Risk Factors for Injury

Body composition is not only an indication of general health; it can also be indicative of fitness and training status in the physically active. Fitness level is an important factor for the physically-active and has been of particular interest in the literature related to the military and athletic arenas because of the implications for performance enhancement and injury reduction (Knapik, Burse, & Vogel, 1983; Malina, 2007). Although the majority of work in this area is not specific to ACL injury, the knowledge gleaned can still be applied to the current context. The following sections will discuss the previous work investigating body composition and strength as risk factors for injury.

Relationship between Body Composition and Injury

Although it has been suggested that body composition should be more closely examined as a risk factor for injury (Griffin, et al., 2006), few studies have examined this relationship. Of those studies, estimates of body composition have been limited to measures of somatotype (Salokun, 1994; Hopper, Hopper, & Elliott, 1995; Hopper, 1997; Lee, Myers, & Garraway, 1997) or body mass index (mass*height^{-2}) (Lee, et al., 1997; Östenberg & Roos, 2000; Knapik et al., 2001; Uhorchak, et al., 2003). Only one of these studies was specific to ACL injury risk (Uhorchak, et al., 2003).

Although not specific to ACL injury, early prospective studies utilized simple methods to classify athletes by somatotype and then observed proportions of injuries that occurred within each cohort (Salokun, 1994; Hopper, et al., 1995; Hopper, 1997; Lee, et al., 1997). In a study of 180 (assumingly male) Nigerian soccer players (Salokun, 1994), subjects classified as ectomorphs (thin) were injured much more frequently than those who were endomorphs (higher proportion of body fat) or mesomorphs (more muscular build). Conversely, another study of 1152 Scottish rugby players (Lee, et al., 1997) found that the ectomorphic players suffered fewer injuries than expected, while the endomorphs had more injuries than were expected. Conversely, a study of 72 top-level female netball players over a 14-week season found that the highest level players with the lower endomorphic profiles were most likely to suffer a lower extremity injury (Hopper, et al., 1995). Although the reason for the disagreement is not entirely clear, these contradictions may speak to the limitations of studying somatotype as a measure of body composition. Although somatotype is gauged by body size and shape to provide a description of physique (Salokun, 1994; Hopper, 1997), this may or may not be related to the magnitudes of fat and lean mass.

A somewhat better estimate of body composition is body mass index (BMI). BMI quantitatively assesses body size by dividing body mass by body height^2, and is reported to relate well to fat mass in the average population (Smalley, et al., 1990; Malina, 2007). However, findings are still equivocal relative to its relationship with injury risk. Östenberg and Roos (Östenberg & Roos, 2000) followed 123 female athletes over the course of one soccer season after an initial screening which included BMI, laxity, and physical performance measures and found no difference in height, weight, BMI, or performance measures in those who went on to suffer a lower extremity injury vs. those who did not. Conversely, in the aforementioned study of Scottish rugby players (Lee, et al., 1997) those who sustained hip or thigh injuries had a significantly higher BMI than those who were not injured. Likewise, in a well-controlled prospective study by Uhorchak et al (Uhorchak, et al., 2003), BMI was a significant predictor of ACL injury risk. In this study, 859 West Point cadets were screened for several anthropometric and laxity variables upon entrance to the academy and then followed over 4 years for the occurrence on an ACL injury. The investigators observed 24 non-contact ACL injuries, of which females suffered 3 times more injuries than the males. Their relative risk analysis revealed that females with a combination of anterior knee laxity and a BMI greater than one standard deviation above the mean were 37.7 times more likely to suffer an ACL injury than those with average values. The risk ratio for high BMI alone was 3.5, second only to a narrow femoral notch width (relative risk= 4.0).

These findings (Uhorchak, et al., 2003) suggest that possessing a higher amount of body fat (and thus less relative muscle mass) may predispose one to injury. However, BMI was not predictive of future injury for males, suggesting that the role of BMI may be different for males and females. Because BMI increases with body mass (i.e. calculated as mass*height^{-2}) without regard for body composition, it remains plausible that elevated BMIs are due to an increase in fat mass in some and an increase in lean mass in others, which may especially be true in an athletic population (Garrido-Chamorro, Sirvent-Belando, Gonzalez-Lorenzo, Martin-Carratala, & Roche, 2009). This point is illustrated by a study of 456 male and female university athletes (Dane, Can, & Karsan, 2002) which found that injured athletes had higher BMI's, but not body fat percentage, than those who did not sustain an injury during the 3 month season. This discrepancy between higher BMI, but not higher body fat, may point to the inability of BMI to accurately assess body composition, which could explain the disparate findings regarding BMI and body type as a risk factor for injury. Specifically, these methods are not sophisticated enough to parse out the contributions of fat mass or lean mass to the increased BMI in the injured group, thus highlighting the need for more specific measures of body composition in our investigations of injury risk. Only one study was found that has used a more accurate assessment of body composition as it relates to injury risk. This study followed 147 males and 138 females through Army basic training for 8 weeks and did not identify BMI, body fat percentage (by DXA), or maximal isometric leg press strength as risk factors for lower extremity injury (Knapik, et al., 2001). However, this study represents a relatively small sample studied over a rather brief period of time.

In summary, epidemiological studies examining body type and BMI as a risk factor for injury remain equivocal, likely due to the questionable ability of these measures to discriminate between lean and fat mass. This may be especially important in light of the maturation-related sex differences in body composition which were previously presented, whereby the increases in body mass in males are primarily due to lean mass, while the primary increase in body mass in females is due to fat mass (Loomba-Albrecht & Styne, 2009). Additionally, it is important to note that of the aforementioned studies, only one specifically addressed the role of body composition (measured as BMI) in ACL injury risk (Uhorchak, et al., 2003). Given the limited research in this area, our understanding of the potential relationship between body composition and injury risk is incomplete and deserves further study. Because body composition is modifiable through training, continued work in this area is particularly relevant to injury prevention.

Relationship between Muscle Strength and Injury

Because the muscles resist the externally-applied torques to stabilize the joints during dynamic motion, this has led to the theory that muscle strength may affect someone's risk of sustaining an injury; specifically, that weaker muscles are less able to perform eccentric work to control external forces and resist joint deformation and injury (Lephart, et al., 2002; Zhang, et al., 2008). However, absolute muscle strength has not been identified as a risk factor for ACL injury per se (Östenberg & Roos, 2000; Knapik, et al., 2001; Uhorchak, et al., 2003). Most often studied is the ratio of knee extensor to flexor thigh muscle strength (Myer, et al., 2009) or dominant to non-dominant thigh muscle strength (Knapik, Bauman, Jones, Harris, & Vaughan, 1991) in order to assess whether muscular imbalances are related to lower extremity injury risk.

Most studies have not found a relationship between muscle strength and lower extremity injury occurrence, whether strength was measured isokinetically (Östenberg & Roos, 2000; Uhorchak, et al., 2003), by field tests known to be correlated with strength (Östenberg & Roos, 2000; Knapik, et al., 2001), or by various strength tests (Jones et al., 1993). In a study of Swedish female soccer players (Östenberg & Roos, 2000), neither isokinetic knee flexor and extensor torque at 60°/s and 180°/s nor vertical or single-leg hop performance were found to be significant predictors of lower extremity injury. Neither isometric leg press nor vertical jump entered into prediction models for lower extremity injury occurrence in male and female soldiers during basic combat training (Knapik, et al., 2001). Likewise, there were no differences in a multi-joint lifting test between injured and non-injured male soldiers in basic training (Jones, et al., 1993). In the prospective 4-year military study previous cited (Uhorchak, et al., 2003), males produced larger concentric and eccentric isokinetic quadriceps or hamstrings torques at 60°/s than females, yet these strength variables did not predict ACL injury for either sex. However, another prospective study has shown that athletes who were injured (including all types of injuries) had less leg power than the non-injured, although their methods for measuring leg power were not well described (Dane, et al., 2002).

Some studies that have reported a relationship between muscle strength and lower extremity injury have found that greater strength and or physical performance measures make one more prone to injury. Hopper et al (Hopper, et al., 1995) found that greater vertical jump performance was actually predictive of lower extremity injury. Since vertical jump test is an oft-used field test which is correlated with leg strength, their findings suggest that possessing greater lower limb strength actually places one at a greater risk for injury, which seems a bit counterintuitive as one would expect better athletic performance (particularly strength and fitness) to be inversely related to injury risk. Östenberg and Roos (Östenberg & Roos, 2000) identified better performance on a multi-directional agility test as a risk factor for injury. They posited that the more elite players, who were found to have a higher risk of injury in their study, performed better on this test, which would support the relation between level of play and injury occurrence which has been observed (Inklaar, 1994; Myklebust, et al., 1997). These findings support the importance of controlling for the skill level of the participants when investigating the direct relation between strength and injury occurrence.

Summary: Body Composition and Strength as Risk Factors for Injury

There is a paucity of literature specifically examining body composition as a risk factor for injury, particularly ACL injury. The literature which has attempted to relate somatotype or BMI to injury is difficult to interpret since these are only crude indicators of relative body fat versus lean mass. Since the consequence of less lean mass is likely reflected in less muscle strength, muscle strength has been investigated to a larger extent. However, this body of literature yields little consensus as well. To date, only one study has sought to examine body composition as a risk factor for ACL injury and it showed that a larger BMI (but not less strength) placed females at a 3.5 times greater risk of future ACL injury (Uhorchak, et al., 2003). This finding is somewhat contradictory since one would likely expect that less strength would also be a risk factor along with a greater BMI. However as previously discussed, BMI is not an accurate predictor of lean mass. In light of this limitation, and findings that females with less lean mass are more likely to sustain an ACL injury, there is a need to further explore the role of body composition on injury risk. These findings also point to the need for a more valid and accurate method of body composition assessment and for a better understanding of how it specifically relates to maximal strength-producing capability of the thigh muscles during dynamic actions, thus their ability to control body movement during deceleration type maneuvers.

Relationships between Body Composition, Strength and Lower Extremity Biomechanics

No studies have directly examined the effect of *in vivo* body composition on landing biomechanics. However, as previously stated, the likely mechanism by which body composition influences neuromechanics is through muscle strength. Specifically, because females possess less relative lower extremity lean mass and also produce lower maximal muscle torques than males, they may have a greater difficulty in controlling the deceleration of their body during landing type movements, which may be reflected in altered neuromechanical strategies. As such, the relationships between muscle strength and biomechanics may provide insight into the effect that body composition has on landing strategies. The following sections will present the current knowledge regarding the influence of strength on lower extremity biomechanics, including the effects of acutely increasing and decreasing strength.

Relationship between Strength and Lower Extremity Biomechanics

Studies comparing thigh strength and landing biomechanics between females and males have consistently shown that, as expected, females possess lower thigh muscle strength than males, as measured by isometric quadriceps and hamstring torques (Shultz, Nguyen, Leonard, et al., 2009; Schmitz & Shultz, 2010) and concentric torques at 60º/s (Lephart, et al., 2002; Salci, et al., 2004). In studies where females produced lower thigh torques than males, they were also observed to demonstrate stiffer landings across different activities. During single-leg landings, females landed with less knee flexion excursion and faster time to peak flexion than their height-matched male counterparts (Lephart, et al., 2002). Similarly, females performed double-leg landings with larger peak knee extensor moments and vertical ground reaction forces than males (Salci, et al., 2004). Females also performed double-leg drop jumps with larger pre- and post- landing muscle activation amplitudes (Shultz, Nguyen, Leonard, et al., 2009) and larger energetic demands at the knee during a drop jump landing (Schmitz & Shultz, 2010) compared to males.

Shultz et al (Shultz, Nguyen, Leonard, et al., 2009) more directly examined the interactive relationships between strength, muscle activation amplitudes, and hip and knee joint biomechanics during a double-leg drop jump task. They first examined the relationship between quadriceps and hamstring MVIC and activation amplitudes, and then the contributions of pre- and post-activation amplitudes to hip and knee kinematics and kinetics once strength was accounted for. They hypothesized that there would be an inverse relationship between quadriceps strength and activation level and that these variables would explain some of the sex differences in landing biomechanics. In this study, maximal isometric thigh torque production was a moderate predictor of peak quadriceps activation amplitude in females, but not in males. However, after controlling for thigh strength and reciprocal hamstring activation, quadriceps activation amplitude was not related to sagittal plane hip and knee motion. Rather, sex (being female) and landing with less hip flexion were the strongest predictors of larger knee extensor moments. Likewise, greater anterior shear force at the knee was predicted to a larger extent by the biomechanical variables, rather than muscle strength or activation. These findings concur with those of Bennett et al (Bennett et al., 2008) who examined the relationship between isokinetic hamstring and quadriceps torques and knee anterior shear force (ASF) in females only during a double leg forward jump landing from a 30cm box. Neither eccentric quadriceps, concentric hamstring contractions (at 60°/s, 180°/s, and 300°/s), nor a ratio of the two muscle groups were significantly correlated with ASF. However, the lack of relationship between peak quadriceps and hamstring torques and biomechanical variables in both studies could be due to the fact that the peak torques are often not produced at the same point in the knee range of motion where the biomechanical variable was produced. For example, Bennett et al (Bennett, et al., 2008) reported that their subjects produced their peak quadriceps and hamstring torques at 60°/s at 73.1° and 31° of knee flexion, respectively, while peak ASF occurred at 37°. It can be assumed that the same issue may have resulted in the non-significant relationships between isometric quadriceps and hamstring torques and ASF reported by Shultz et al (Shultz, Nguyen, Leonard, et al., 2009).

To address this issue, Schmitz et al (Schmitz & Shultz, 2010) compared energy absorption strategies between males and females during a double-leg drop jump landing and examined the extent to which quadriceps and hamstring maximal isometric torques could predict lower extremity energy absorption. Lower extremity energy absorption (to be discussed in more detail in the section to follow), is calculated over the entire deceleration phase, and represents the eccentric work of the musculature, which may result in more meaningful relationships between strength and landing mechanics than a single biomechanical value at a specific point in time (e.g. peak anterior knee shear force). Their findings revealed that females were significantly weaker than the males in both muscle groups, but absorbed a similar amount of total energy. However, the females demonstrated a distinct pattern by which they utilized 69% more knee flexion range of motion, thus resulting in a greater relative proportion of energy absorbed at the knee. While neither quadriceps nor hamstring MVIC torques were significant predictors of energy absorption for males, greater quadriceps MVIC predicted greater energy absorption at the knee in females, which suggests that those females who possessed greater knee extensor strength were better able to control the deceleration phase of the landing, thus providing the time and range of motion needed to allow for energy dissipation. The reason for this sex-specific relationship between strength and energy absorption was not clear. The authors posited that the relationship may have been due to unequal magnitudes of task difficulty by which the drop jump required a relatively greater proportion of the females' maximal isometric strength compared to the stronger males. This is an important observation especially in light of our knowledge of sex differences in body composition. That is, since females possess less lean mass relative to their total body mass compared to males, it will be relatively more difficult for them to perform an equivalent task (e.g. 0.45m drop jump), since they have less strength to control the same amount of relative body mass compared to males. Hence, it is possible that the increased relative difficulty of this task due to the lower amount of lean mass in females may be influencing their neuromechanical strategies.

Effect of Training-Induced Increases in Strength

If a lack of strength and thus inability to efficiently control landing (Lephart, et al., 2002) is indeed the driving force behind these unsafe neuromechanical strategies, then one might expect that increasing strength would result in a landing pattern with more controlled hip and knee flexion, thus enabling the muscles of the hip and thigh to absorb more energy. Although strength-related improvements in energy absorption have not been examined, one study of 11 female high school volleyball players who completed a 6-week plyometric training program showed an increase in isokinetic (but not isometric) hamstring torques at 360°/s along with a concurrent 50% decrease in vertical ground reaction forces during a vertical jump landing (Hewett, et al., 1996). These findings suggest that increasing strength leads to improvements in landing mechanics. Interestingly, this group also tested isometric hamstring torque, but the subjects did not show improvement following the training program, which suggests that the mode by which muscle strength is tested may have significant implications for our ability to interpret relationships between strength and landing mechanics. Illustrating this point, another study showed significant increases in MVIC strength of the hip and knee in 33 female college-age recreational athletes following a training program, but no changes in lower extremity kinematic or kinetic variables for any lower extremity joint during a stop-jump task (Herman et al., 2008). These conflicting results are likely a result of differences in strength training (typical isotonic weight training exercises (Hewett, et al., 1996) vs. slow contractions with a resistance band (Herman, et al., 2008)), type of contraction tested (isokinetic (Hewett, et al., 1996) vs. isometric (Herman, et al., 2008)), length of training program (2h sessions, 3x/wk for 6 weeks (Hewett, et al., 1996) vs. 3x/wk for 9 weeks (Herman, et al., 2008)), and landing task (vertical jump landing (Hewett, et al., 1996) vs. primarily horizontal stop-jump (Herman, et al., 2008)).

Because of the limited interpretability of these studies due to methodological differences, we do not have a complete understanding of how increasing strength affects lower extremity neuromechanics. However, the discrepancy between these findings has highlighted the inherent difficulty with isolating the effects of increased muscle strength from other neuromuscular factors when evaluating improvements in landing mechanics. As such, investigating the effect of increased strength on landing mechanics may not be the optimal way of initially evaluating the relationship between strength and lower extremity neuromechanics.

Effect of Artificial Decreases in Relative Strength

The effect of strength changes on lower extremity neuromechanics has also been examined via an acute increase in body mass without a concomitant increase in lean mass. Three recent studies of recreationally-active college co-eds (C. N. Brown, et al., 2005; Kulas, et al., 2008; Kulas, et al., 2010) have provided insight into these effects by artificially adding 10% of body mass to each participant's trunk, thus effectively reducing their strength relative to their total body mass. Compared to the un-weighted condition, the additional mass condition resulted in a more erect landing position and larger peak knee extensor moments during a stop-jump task (C. N. Brown, et al., 2005), and larger ground reaction impulses, knee extensor impulses, and increased energy absorption at the knee and ankle during a double-leg drop landing (Kulas, et al., 2008). Additionally, the added weight resulted in larger modeled quadriceps and gastrocnemius forces (Kulas, et al., 2010). While the findings from these studies were consistent across participants, neither study performed separate analyses for males and females. This is a key limitation, as females already possessed lower relative strength compared to males in the unloaded condition. Therefore, adding the same relative amount of mass to males and females (10% BW) may have systematically increased the difference in relative strength to total body mass, thereby placing relatively larger demands on the females who already possess less strength to total body mass to begin with.

These studies provide preliminary evidence that reducing relative strength (by artificially adding mass to the body without a compensatory gain in lower extremity muscle mass) may result in greater demands on the knee extensors to decelerate the body

(C. N. Brown, et al., 2005; Kulas, et al., 2008; Kulas, et al., 2010) and landing with a more erect landing posture (C. N. Brown, et al., 2005), both of which have been linked to increased strain on the ACL (Decker, et al., 2003; Hewett, et al., 2005; Yu, et al., 2006; Blackburn & Padua, 2009). Since it is known that females have less relative lower extremity strength compared to males, this suggests that the decreased amount of lower extremity lean mass and strength in females may in part be dictating this at-risk landing strategy more often observed in females.

Summary: Relationships between Strength and Lower Extremity Biomechanics

The existing body of literature indicates that the lower amounts of strength in females may in part explain the stiffer landings performed with less hip and knee flexion (Lephart, et al., 2002), greater demands on the knee extensors (Salci, et al., 2004; Kulas, et al., 2008; Shultz, Nguyen, Leonard, et al., 2009; Kulas, et al., 2010; Schmitz & Shultz, 2010), and greater landing forces (Salci, et al., 2004) observed in females compared to their stronger male counterparts. While it is not quite clear that increasing strength alone improves landing mechanics, it does appear that artificially decreasing strength results in performing landing maneuvers in a manner that is thought to place strain on the ACL (C. N. Brown, et al., 2005; Kulas, et al., 2008; Kulas, et al., 2010). Applied to the current context, these findings lend support to the theoretical framework that the at-risk neuromechanical strategies used by females may be a consequence of their decreased muscle mass to total body mass (thus decreased relative strength) compared to males.

Section Summary:

Potential Relationships between Body Composition, Lower Extremity Biomechanics, and Injury

There is currently no consensus in the literature regarding the role of body composition on lower extremity injury. However, studies to date have been limited to the use of BMI and somatotype as measures of body composition, which do not adequately account for the specific contributions of lean versus fat mass to body size. In the only study to date examining the effects of body composition on ACL injury, females with an above average BMI were reported to be at 3.5 times greater risk of sustaining an ACL injury (Uhorchak, et al., 2003). Thus, it is important to continue work in this area using appropriate measurement methods to clarify the role of body composition on injury risk.

This section primarily focused on the relationship between strength and lower extremity neuromechanical strategies because it is most likely that the mechanism by which body composition affects injury risk is through strength-related influences on neuromechanical strategies. Females possess less relative lower extremity lean mass and also produce lower maximal muscle torques than males, which likely leads to a greater difficulty in controlling the deceleration of the body during maneuvers such as landing. This is supported by studies that found that females who exhibited lower amounts of knee flexor and extensor strength also exhibited stiffer landings performed with less hip and knee flexion, greater peak knee extensor moments and greater demands on the quadriceps to absorb energy compared to their stronger male counterparts (Hewett, et al., 1996;

Lephart, et al., 2002; Salci, et al., 2004; Shultz, Nguyen, Leonard, et al., 2009; Schmitz & Shultz, 2010). These potential associations are further demonstrated when artificial decreases in strength induce landing strategies that are thought to place increased strain on the ACL (C. N. Brown, et al., 2005; Kulas, et al., 2008; Kulas, et al., 2010). Collectively, these findings suggest that the neuromechanical strategies used by females may be a consequence of their decreased strength compared to males.

There are several limitations identified in this body of literature. An emerging theme from this review is the difficulty in examining the relationships between peak muscle torques produced at one point in the range of knee motion and a peak neuromechanical variable produced at another point in the range of motion. Also important is the inconsistent muscle contraction type which has been used to examine strength, as all strength testing modes may not appropriately represent the required muscle actions of the functional activity. In order to further elucidate the role of strength on landing neuromechanics, it may be prudent to consider more specific muscle strength measurements as they relate to a more global measure of lower extremity biomechanics. The next section will discuss these important methodological considerations to best address the limitations of the current literature in order to effectively determine the extent to which body composition and strength influence lower extremity neuromechanical strategies.

Methodological Considerations

This literature review thus far has attempted to lay a foundation for a plausible connection between sex differences in body composition and strength with sex differences in lower extremity neuromechanics as they relate to the increased risk of ACL injury in females. Because these variables have not been examined collectively, review of the constituent parts indicates that there are several methodological considerations that must be accounted for in order to effectively examine this combination of variables and the relationships that may exist among them. The following section will present important methodological considerations for body composition, strength, and biomechanical assessment to support the most appropriate methods for each measurement for the proposed work.

Body Composition Assessment

Because lean muscle mass is a main determinant of the ability to create and absorb forces, the ability to accurately measure body composition is crucial in order elucidate the role of sex differences in lean mass on sex differences in neuromechanical strategies. Although extensively used in the literature, rudimentary measures of body composition such as BMI are not sensitive enough to discriminate between increases in body mass due to fat mass versus lean mass. More accurate and commonly-used clinical methods of measuring body composition are available to the clinician and researcher alike, include hydrostatic weighing and anthropometric measurements. Additionally, mounting literature supports the use of dual-energy x-ray absorptiometry (DXA) for assessing body composition. Each method requires varying degrees of invasiveness to the subject and the skill required of the clinician. The following section addresses the strengths and weaknesses of the most commonly used methods for estimating or assessing body composition in the sports medicine literature.

Body Mass Index

The Quetelet Index, or as it is more commonly known, Body Mass Index (Eknoyan, 2008), is a commonly-used tool to assess an individual's general degree of fatness and is calculated by dividing body mass (kg) by height^2 (m). While not an actual measurement of body composition, BMI is used frequently because it is simple to measure and provides a reasonable estimate of body fat percentage in the general population (Knapik, et al., 1983; Smalley, et al., 1990; Malina, 2007; Garrido-Chamorro, et al., 2009). Hence, most investigators have utilized this measure as a screening variable in their prospective studies as a representation of the degree of fatness of their subjects. While one study reported that BMI is highly correlated ($r=\sim0.80$) with body fat percentage (measured by DXA) in adult females, explaining 65.9% of the variance in body fat percentage, they also noted that body fat percentage could change ±5% without a concurrent change in BMI (Hannan, Wrate, Cowen, & Freeman, 1995). Further, the 95% confidence interval for body fat percentage in females whose BMI was 20.0 kg/m^2 ranged from 13.1%-31.5%, which represents a seemingly large clinical range in body composition. While the limitations of BMI are evident in the general population, the ease of measurement and its ability to predict various obesity-related pathologies (Garrido-

Chamorro, et al., 2009) perpetuate its use. However, these limitations may be more pronounced, and therefore of larger concern when studying an athletic population. Nevill et al (Nevill, Stewart, Olds, & Holder, 2006) compared sum of skinfolds (to estimate body fat) and BMI between male and female athletes and non-athletes, and between different sports within the same sex. They reported that BMI was more highly related to body fat in females (r=0.72) vs. males (r=0.48). Moreover, the relationship between BMI and sum of skinfolds was different between different types of athletes. These findings indicate that BMI is not sufficiently accurate to provide a similar measurement of adiposity between (athlete vs. non-athlete, male vs. female) or within (male athletes, female athletes) populations.

The inconsistencies when using BMI to estimate fatness in an athletic population are largely attributed to the larger percentage of body mass composed of muscle mass (Nevill, et al., 2006; Ode, Pivarnik, Reeves, & Knous, 2007; Garrido-Chamorro, et al., 2009). In fact, a study of college athletes found that 73% of males and 34% of females who were within the normal range for body fat percentage (as measured by air displacement plethysmography) were misclassified as overweight by BMI (Ode, et al., 2007). In a large study of national-level athletes, primarily soccer and basketball players, BMI was highly related to body fat percentage in females (r=0.76) but was only moderately correlated for males (r=0.61) and lacked the ability to predict percent body fat (Garrido-Chamorro, et al., 2009).

Collectively, these studies demonstrate that while BMI may be used to get a general idea of body fatness, it is not an accurate measure of body composition if one wishes to make comparisons across sex or within athletic populations. This limitation may explain the contradictory findings in epidemiological studies that used BMI to assess injury risk, or why BMI was a stronger predictor of ACL injury risk in females compared to males. Thus, more accurate methods of assessing body composition are necessary to elucidate the role of relative lean muscle versus fat mass on landing strategies and risk for injury.

Whole Body Composition Assessment Methods

The two most frequently used methods for assessing body composition in sports medicine applications are hydrodensitometry and anthropometry. Hydrodensitometry, also known as "underwater weighing" or "hydrostatic weighing" works on the principle that the density of specific body tissues are known and constant (Malina, 2007) and that when completely submerged in water, body volume is equal to that of the water which it displaces after correcting for the density of the water (Going, 2005). Considered the "gold standard" of body composition in sport science applications, this method allows for calculation of body density which is then used in an equation to predict fat and fat-free mass. Because of inter-individual differences in fat-free mass composition and density between populations, the Siri 3-C model is most commonly-used and is considered the most valid prediction equation (Going, 2005). However, limitations to hydrodensitometric measurement include the technical skill needed to perform the measurements and the difficulty for the subject to 1) completely submerge their body underwater, and 2) maximally exhale and then hold their breath while the tester makes an accurate reading of the scale. Another major limitation is that hydrostatic weighing cannot be used to estimate regional body composition.

Anthropometry, on the other hand, involves taking simple measurements of limb circumference and skinfold thickness and offers the most portable and inexpensive method of whole and regional body composition assessment. In order to assess whole body composition, skinfold thickness must be measured at several sites, and the sum of those skinfolds are then used in a standard formula (Bellisari & Roche, 2005). Skinfold thickness and limb circumference can also be used to estimate limb cross-sectional area (bone + lean mass) (Wang et al., 1999; Bellisari & Roche, 2005); however this method is problematic, as these results are influenced by inter-individual differences in skin thickness and compressibility and are also non-specific since they include skin, muscle, bone, adipose tissue, etc. (Bellisari & Roche, 2005). As such, radiographic methods such as computed tomography (CT), magnetic resonance imaging (MRI), and dual-energy x-ray absorptiometry (DXA) are commonly accepted as more valid means for predicting regional lean mass (Bellisari & Roche, 2005; Lukaski, 2005).

Regional Body Composition Assessment Methods

Radiographic methods for assessing regional body composition, specifically lean mass, have emerged as important tools for clinicians and researchers. Computed Tomography (CT), Magnetic Resonance Imaging (MRI), and Dual-energy X-Ray Absorptiometry (DXA) allow for direct visualization and quantification of the tissues present within the region of interest (ROI) (Lukaski, 2005). While both CT and MRI provide high-resolution 3-dimensional images to quantify amounts of fat, lean, and bone, and are considered the gold standard for regional body composition assessment (Wang, et al., 1999; Plank, 2005; Hansen et al., 2007), the use of these methods is limited by the high cost of the equipment and operation. Additionally, exposure to higher levels of radiation (CT) and long scan times (MRI) limit their practical use in various populations (Lukaski, 2005). Because of these limitations, DXA has emerged as the preferred method for both whole and regional body composition assessment.

Dual Energy X-Ray Absorptiometry

Recently, Dual Energy X-ray Absorptiometry (DXA) has been introduced as a more practical tool to measure body composition and is now widely used for total and regional body composition assessment (Plank, 2005; Andreoli, Scalzo, Masala, Tarantino, & Guglielmi, 2009). The ability of DXA to accurately assess total body and regional lean and fat mass is well accepted and is being promoted as an important clinical tool for the sport sciences (Andreoli, et al., 2009).

DXA uses a three-compartment (bone, fat-free, and fat) model and is accepted as an accurate method for determining total and regional body composition (Andreoli, et al., 2009). During a whole body DXA scan, the patient lays flat (usually supine) on a table, under which lies an X-ray emitter. During the scan, x-ray beams are directed anteriorly through the body at two different energies. These beams are then attenuated as they hit the bodily tissues, which absorb or scatter the beam (Pietrobelli, Formica, Wang, & Heymsfield, 1996). The beam detector which is located in the scanning arm, travels down the body rectilinearly (from proximal to distal), in “sweeps”. During each sweep, the detector compares the attenuated beams with the original beam, based on the known attenuation of the beam as it travels through the different tissues. The software then uses a “gradient approach” to find areas of large attenuation to identify the bone-soft tissue interface (Watts, 2004) and then divides each pixel into its constituent compartments (Pietrobelli, et al., 1996). However, only two components can be distinguished within any pixel (Andreoli, et al., 2009). Since pixels that contain bone and soft tissue essentially have three tissue compartments present (bone, fat, lean), an extrapolation technique is used to estimate the composition of any pixel that contains bone (Andreoli, et al., 2009). A major assumption of DXA is that the composition of pixels which cannot be measured (because they contain bone) are equal to those that can be measured (because they only contain fat and lean tissue) (Lohman & Chen, 2005). Since approximately 40% to 45% of a whole body scan contains bone and soft tissue pixels, a systematic individual error may be introduced due to differences in the composition of measured pixels versus the estimates assigned to the non-measured areas (Lohman & Chen, 2005). This error is thought to particularly affect body composition estimates in the thorax and arms because of the relatively large areas of bone in those areas. However, it is difficult to assess the amount of error since the extrapolation technique is specific to each manufacturer’s proprietary algorithms (Pietrobelli, et al., 1996; Lohman & Chen, 2005).

Fortunately, it does not appear that this extrapolation error affects the lower extremity to the same extent. Several validation studies have been performed which show that DXA can produce measurements of lower limb fat and lean mass which are highly related to those of CT (R^2=0.73-0.96) (Visser, Fuerst, Lang, Salamone, & Harris, 1999;

Wang, et al., 1999; Levine et al., 2000) and MRI (r=0.93-0.98) (Fuller et al., 1999; Elia et al., 2000). As mentioned previously, these methods are not typically preferred when considering the expense, risks to the patient, and the relative unavailability to the clinician (Plank, 2005). Hence, DXA has become an attractive alternative for body composition assessment because it can be performed without much of the cost and risk associated with the more sophisticated instruments (Andreoli, et al., 2009).

Standard procedures ensure the reliability of the measurements and analysis of the images from DXA. The DXA hardware and software are calibrated daily using a phantom comprised of known densities. An acceptable coefficient of variation is necessary before performing a scan, thus ensuring the day-to-day mechanical reliability of the machine itself. Software provided by the DXA manufacturer then offers the ability to create standard and custom regions of interest (ROI) in order to analyze the composition of any segment desired. This is achieved by manually setting the boundaries of each ROI using pre-determined anatomical landmarks. Reliability of manual ROI placement in the lower extremity in 100 college-aged males and females (mean age= 21.8±6.2 years, BMI=24.0±3.53 kg/m^2) was reported to range from 0.998 to 0.999 and 0.995 to 1.000 for fat and fat-free mass, respectively (Burkhart, Arthurs, & Andrews, 2009). One limitation however that has been identified with the use of DXA as opposed to MRI or CT for regional assessment is the inability to separate out the lean mass of particular muscle groups (Bamman, et al., 2000). MRI and CT provide pictures of cross-sectional slices of the limb where the individual muscle groups can be traced and area calculated whereas DXA calculates all lean mass of a limb segment. As such, correlations between muscle mass and knee extension torque, for example, will include the mass of all thigh musculature not just the quadriceps. This limitation is illustrated by the results from a study of 39 women which showed that maximal voluntary isometric contractions of the plantar flexor muscle group were significantly correlated to various measures of lower leg lean mass (range: r= 0.365-0.733), but that those measures that included only the plantar flexors explained much more of the variance (42.2-53.7%) in MVIC than the measures that included all lean mass of the lower limb (13.3-14.6%) (Bamman, et al., 2000). While this limitation exists, the advantage of getting a closer estimate of the available contractile mass in an efficient and non-invasive manner may still prevail.

Summary: Body Composition Assessment

While several methods exist, the advantages and limitations of each must be weighed in order to determine the most appropriate method for assessing body composition. Standard methods such as anthropometry and hydrodensitometry offer well-accepted measures of whole body composition, specifically percent body fat. However, hydrodensitometry lacks the ability to make regional measures of lean mass and the ability of anthropometry is questionable. Computed Tomography and Magnetic Resonance Imaging offer accurate measures of regional lean mass, but are often unavailable or impractical in a research setting. As such, Dual-energy X-ray Absorptiometry has emerged as a valid and reliable method for measuring regional lean mass and has many applications for the sports medicine researcher and clinician alike.

Strength Assessment

Some of the aforementioned discrepancies in the relationship between strength and landing mechanics may be due to differences in methods by which muscle strength is assessed. It is well accepted that muscle strength follows a principle of specificity, not generality, whereby muscle strength measures which most closely mimic specific dynamic actions will be more closely related than when a static (i.e. isometric) measurement is used (Baker, Wilson, & Carlyon, 1994). As such, it may be necessary to measure strength in a manner that is most consistent with the muscle action performed during the experimental task. During landing, the knee extensors are lengthened as they contract to produce a controlled deceleration of the body. Hence, measuring eccentric strength of the knee extensors would be most relevant to the task. However, there is some question as to which mode of strength testing will best represent the function of the hamstrings during landing (Bennett, et al., 2008). This is because of the bi-articular nature of the hamstring muscles whereby they are shortened distally and lengthened proximally as the knee and hip are flexed during landing, respectively (Withrow, Huston, Wojtys, & Ashton-Miller, 2008; Powers, 2010).

Although concentric contractions have been used frequently to assess hamstring strength during biomechanical studies (Hewett, et al., 1996; Lephart, et al., 2002; Salci, et al., 2004; Bennett, et al., 2008), rarely is a rationale provided. This is most likely due to the disagreement as to the primary action of the hamstrings during landing. However, there is some evidence that eccentric hamstring action is an important factor in resisting anterior tibial translation during landing. In a cadaveric study, Withrow et al (Withrow, et al., 2008) simulated a drop landing during isotonic, lengthening, and shortening contractions of the hamstrings. When they applied a 1700 N compressive impulse to cadaveric knees, the simulated eccentric hamstring forces resulted in lower peak ACL strain compared to the conditions where they applied isotonic or shortening hamstring contractions. This suggests that during landing, eccentric hamstring torques are more effective at reducing ACL strain than isometric or concentric contractions. Accordingly, the capacity to apply that torque (i.e. maximal strength) may be an important determinant in reducing ACL strain and risk of injury and thus merits further study.

An additional factor to consider is the outcome variable to which strength is being related. In the current context, the outcome of interest is energy absorption which is reflective of the amount of eccentric work which is performed during landing. Since the hamstrings work eccentrically at the hip to control hip flexion and forward motion of the trunk during landing (Powers, 2010), eccentric hamstring torque is more consistent with the measure of hip energy absorption and/or torsional stiffness.

Biomechanical Assessment

During landing, females are reported to demonstrate an upright landing posture, larger knee extensor moments, faster time to peak flexion, and larger vertical ground reaction forces, altogether described as a “stiff” landing (Devita & Skelly, 1992; Lephart, et al., 2002; Decker, et al., 2003; Schmitz, et al., 2007). However, since it appears that the body has different methods by which it can modulate the lower extremity to resist the external forces imposed during landing (Hewett, et al., 1996), the aforementioned components of a stiff landing are not always observed together. For example, Yu et al (Yu, et al., 2006) observed decreased hip and knee angles in females at ground contact, but no sex difference in vGRF or anterior shear force. Likewise Hewett et al (Hewett, et al., 1996) observed decreased landing forces following a training program, but no improvement in peak knee extensor or flexor moments. Because of the complex interaction between kinematic and kinetic variables, examining energy absorption patterns as an indication of the "global strategy" utilized to decelerate the body (McNitt-Gray, 1993; Schmitz, et al., 2007) has been suggested because it takes into account joint position, moments, and the time over which this action occurs. This section will discuss the theoretical basis of energetics and its relationship to injury, and how changes in landing stiffness are expressed.

Energy Absorption

Energy absorption is calculated by integrating the negative portion of the power curve (product of joint moment and angular velocity) and describes the negative work or eccentric action of the lower extremity muscles during the deceleration phase of a landing (Devita & Skelly, 1992; McNitt-Gray, 1993; Schmitz, et al., 2007; Zhang, et al., 2008; Schmitz & Shultz, 2010). Absorption of the kinetic energy generated during landing by the muscles is thought to be advantageous over energy absorption of the passive structures (bone, cartilage and ligaments), which may result in both acute and chronic injuries (McNitt-Gray, 1991; Zhang, et al., 2000; Zhang, et al., 2008). However, because there are no studies to date which have examined the relationship between energy absorption and injury occurrence, this relationship remains theoretical.

The theory underlying the study of energetics in sports medicine and biomechanics research is related to the fundamental physical principle that energy is neither created nor destroyed; it only changes forms. In the current context, when a person is standing at height above the ground and is not moving, they possess an amount of stored energy ("potential energy"), which can be calculated as the product of body mass (kilograms), acceleration due to gravity (g) (m/s^2), and their distance (height) above the ground (h) (meters). When they fall from this height (as in a drop landing), their potential energy is then converted into kinetic energy, which is equal to ½ mass (m) (kilograms) x velocity upon landing^2 (v) (m/s^2) (Devita & Skelly, 1992). Per the Law of Conservation of Energy, the potential energy of the person standing on the top of the box is theoretically equal to the kinetic energy that they generate upon landing. This kinetic energy (usually expressed as ground reaction force) is what creates the external forces around the joints, and which must be absorbed by the musculature in order to produce a controlled deceleration through joint flexion, and subsequent arrest of downward motion (Devita & Skelly, 1992). It is theorized that when there is a failure of the musculature to produce the necessary eccentric torques to absorb/dissipate that kinetic energy that the energy is transferred to other structures, particularly the bony, cartilaginous, and ligamentous structures, thus resulting in injury, both acutely and over time (Zhang, et al., 2000).

Effect of Landing Stiffness on Energy Absorption

Studies have investigated energy absorption patterns during soft vs. stiff landings (Devita & Skelly, 1992; Zhang, et al., 2000), and these studies support the hypothesis that stiffer landings expose the body to greater vertical ground reaction forces, which if not properly attenuated, expose the passive structures to more energy. In a study of 8 female volleyball players who were asked to land from a 59cm height in a soft and stiff manner (90°>knee flexion>90°, respectively), Devita and Skelly (Devita & Skelly, 1992) showed that the stiffer landing resulted in larger vertical ground reaction forces and significantly less energy absorption at the hip and knee, while the ankle absorbed significantly more energy. These findings agree with a similar study where males were asked to land in either a soft, normal, or stiff manner (Zhang, et al., 2000). These authors reported that as the landings became stiffer, vertical ground reaction forces increased, hip and knee flexion excursions decreased and knee extensor moments increased, altogether resulting in larger proportions of energy absorbed about the ankle with a concurrent decrease in absorption at the hip. The increase in energy absorption at the ankle during stiff landings is most likely the result of a stiffer landing being performed in a more erect position, thus reducing the abilities of the hip and knee to absorb energy during deceleration through controlled flexion (Zhang, et al., 2000). These findings mirror those which report that females tend to absorb more energy about the ankle and knee, with less at the hip, compared to males who land more softly (Decker, et al., 2003; Schmitz, et al., 2007; Schmitz & Shultz, 2010).

Summary: Biomechanical Assessment

Females typically perform stiff landings, which are characterized by various combinations of kinematic and kinetic features. Stiff landings may be indicative of a lessened ability to decelerate their total body mass through controlled joint flexion due to a lower amount of lean mass relative to body mass (Lephart, et al., 2002). Measuring energy absorption amalgamates these biomechanical variables and is indicative of the amount of eccentric work performed by the available lower extremity lean mass throughout the entire deceleration phase. Hence, this method offers the most integrative measure to examine the relationship between lower extremity lean mass and neuromechanical strategies during a vertical landing task.

Equalization of Task Demands

The literature has demonstrated that the relationships between strength and landing biomechanics seem to be more pronounced in females than males (Shultz, Nguyen, Leonard, et al., 2009; Schmitz & Shultz, 2010), suggesting that strength is a more critical factor in controlling landing for females. These investigations have typically compared males and females during an equivalent task (i.e. drop landing from a standard height) without regard for inter-subject differences in body composition or physical ability. This could result in the task being relatively more difficult for females (who have less muscle mass available to decelerate their total body mass) compared to males (who possess a larger proportion of lean mass). Occasionally, investigators will scale the task demands to account for inter-subject size differences (e.g. a forward hop equal to a percentage of body height) (Lephart, et al., 2002; Hanson, et al., 2008). However, since males are still relatively stronger than females after adjusting for differences in body size, this adjustment may not be adequate to account for differences in relative strength.

Investigating sex differences in neuromechanical strategies under equalized task demands may help remove this bias and elucidate the extent to which sex differences in body composition underlie the sex differences in knee joint neuromechanics. As discussed previously, the relationship between lower extremity lean mass and neuromechanical strategies during a vertical landing task may be best evaluated through observation of energy absorption strategies. In this context, one methodological choice is to equalize the task demands based on the available lower extremity lean mass relative to the amount of energy which must be dissipated upon landing from a height.

While no studies to date have specifically attempted to equalize task demands by manipulating the experimental task, two studies (James, Bates, & Dufek, 2003; Swartz, Decoster, Russell, & Croce, 2005) have provided a similar rationale for taking into account the differences in task demands due to differences in body mass, landing momentum, and drop height resulting from inter-subject variation in maximal vertical jump height. These normalization procedures ensured that they were accurately comparing the landing strategies based on the differences in physical demands due to differences in body mass and drop height. Interestingly, Swartz et al (Swartz, et al., 2005) found no sex differences in hip and knee kinematics when adolescent and adult participants performed a vertical jump equal to 50% of their maximal ability. Additionally, there were no sex differences in peak vertical ground reaction forces when they were normalized to each individual's kinetic energy. These findings support the contention that a greater parity in task demands may result in more similar landing strategies than what has previously been reported.

In the current context, instead of simply *accounting* for differences in task demands through normalization procedures, the relative task demands can be *controlled* by individualizing the task for each participant. In terms of Newtonian principles, in order to alter the energy upon landing, there must be a change in either body mass, height of the fall, or gravitational acceleration. Two studies (C. N. Brown, et al., 2005; Kulas, et al., 2008) that manipulated body mass showed that artificially increasing body mass induced dangerous landing strategies. However it is not clear whether some of those changes were due to the acute addition of that mass to the trunk versus a more uniform distribution and adaptation over time expected with physiological weight gain. Thus, a better option may be to manipulate the height from which they drop, thus controlling the amount of energy that must be dissipated upon landing. This way, accurate comparisons can be made between males and females by ensuring that the amount of lower extremity lean mass available to dissipate those landing forces is equalized between males and females. In turn, this will lead to a greater understanding of how body composition and thus relative strength influence movement strategies during dynamic tasks where ACL injury typically occurs.

Section Summary: Methodological Considerations

Several methods exist for examining body composition, strength, and lower extremity neuromechanics during dynamic tasks. Although whole body composition methods have been traditionally used in the sport sciences, DXA now provides a valid and reliable method for determining whole and regional body composition. Specifically, this method allows for the measurement of the lower extremity lean mass which is responsible for providing dynamic stability to the knee. Since ACL injuries typically occur during rapid deceleration maneuvers, it appears that the optimal methods for relating muscle strength to lower extremity biomechanics during deceleration would be to measure eccentric strength as an indicator of the maximal capacity to perform eccentric work. That is, muscle energy absorption. Lastly, ensuring that the experimental task places equal demands on males and females can ensure that observed sex differences in neuromechanics are not the result of females performing a more difficult task relative to their abilities. Together, this approach will optimize our ability to evaluate the relationship between lean mass and neuromechanical strategies during deceleration activities where ACL injuries typically occur.

Chapter Summary

The purpose of this literature review was to present body composition as a relevant factor underlying the sex-specific neuromechanical strategies which are thought to place females at a higher risk for ACL injury. While body composition has rarely been examined as a risk factor for ACL injury, it is likely that there are intermediary factors such as strength and lower extremity neuromechanics involved in this relationship. Accordingly, this review examined the potential relationships among body composition, strength, lower extremity neuromechanics and injury. Several key points have emerged which support these relationships:

1) The well-documented sex differences in lower extremity biomechanics suggest that there are unidentified sex-dependent factors which may cause females to perform high-risk deceleration activities in a stiffer manner, with large contributions from the knee extensors and larger ground reaction forces, which are thought to contribute to ACL strain. The potential relationship between strength and these landing strategies has been suggested, but work is incomplete to establish this relationship.

2) Significant differences in total and regional extremity body composition emerge between males and females following puberty, and are highlighted by males possessing significantly more lean mass, less fat mass, and greater strength relative to total body mass compared to females.

3) Possessing an above average BMI may place females, but not males, at an increased risk of sustaining an ACL injury. This suggests that higher fat mass, and thus less lean mass may increase the propensity for females to demonstrate unsafe landing mechanics leading to injury.

4) Preliminary work which manipulated the body composition and relative strength capabilities of both males and females induced stiffer landings and greater reliance on the knee extensors for deceleration, such as those typically observed in females. This suggests that possessing less strength relative to total body mass may be a

driving factor in neuromechanical strategies which are thought to increase ACL strain and result in injury.

Although collective examination of the literature encapsulating these areas suggests that body composition could be a relevant factor underlying at-risk movement strategies observed more often in females, this conjecture has not be examined specifically. Moreover, the existing literature has overwhelmingly compared sex-specific landing mechanics during tasks that are presumably more difficult for females since they possess less lean mass and strength than males. As such, we do not know which sex-specific movement strategies are the result of unequal task demands, nor do we understand how lower extremity lean mass (regardless of sex) influences biomechanics which are thought to strain the ACL and lead to injury. Therefore, work is now needed to examine the influence on body composition (thus lower extremity strength relative to body mass) on lower extremity neuromechanics during an equalized landing task requiring deceleration of the body's momentum, the type of activity commonly observed at the time of injury.

CHAPTER III

RESEARCH DESIGN AND METHODS

Objective

The primary objective of this study was to determine the extent to which sex differences in lower extremity lean mass (LELM) relative to total body mass explain sex differences in energy absorption strategies during a drop jump landing task. To achieve this objective, males and females paired by similar body mass index (BMI), completed body composition and eccentric thigh strength testing before lower extremity energetics were measured during a drop jump landing task from a standard height of 0.45 meters. The relative task demands for the matched pairs from this height were compared by calculating the amount of LELM relative to the amount of potential energy ($LELM*PE^{-1}$) to be dissipated upon landing from the 0.45 meter height. In order to further explore the effects of task demands based on the differences in available lean mass within each matched male and female pair, the drop height was increased to match the male's $LELM*PE^{-1}$ ratio with his female counterpart at the 0.45m height ($Height_{STD}$). Both the male and female within each matched pair performed the drop jump from this second height ($Height_{EQU}$). Energy absorption was calculated from the 3D kinematic and kinetic data during the landing tasks, while LELM was measured by dual energy x-ray absorptiometry (DXA). From these data, the extent to which lower extremity lean mass

(LELM) and peak eccentric quadriceps ($Quad_{ECC}$) and eccentric hamstring (Ham_{ECC}) strength predict energy absorption at the hip, knee and ankle (EA_{HIP}, EA_{KNEE}, and EA_{ANK}, respectively) during the landing phase of the drop jump task was determined within sex. Additionally, males and females were compared on energy absorption with and without LELM*PE^{-1} equalized.

The central hypothesis was that less lean mass relative to body mass would be accompanied by a lower amount of eccentric strength relative to body mass, which would predict less energy absorption in females when landing from the standardized drop height. Further, it was hypothesized that sex differences in landing energetics would diminish once sex differences in LELM-PE^{-1} were equalized, but would be exploited at higher task demands. This was based on previous findings that lower muscle strength is related to greater muscle activation and less energy absorption at the knee and that artificially adding mass (and thus reducing relative strength) results in a more erect landing position and greater knee extensor moments. The rationale for examining the influence of body composition and lower extremity strength on landing biomechanics was that the literature indicates that an above average body mass index in females places them at a higher risk of non-contact ACL injuries and that artificially increasing body mass (thus decreasing relative strength) promotes neuromechanical strategies thought to place strain on the ACL and place one at risk for injury. Further, by equalizing task demands to the differences in body composition in matched males and female pairs, it would be possible to separate out the effects of sex and muscle strength on sex differences in movement strategies. Ultimately, the goal for this research is to identify a body composition profile (lean tissue composition and distribution) which places an individual at an increased risk for injury and to create appropriate intervention strategies to that end.

Subjects

Seventy males and females (35 males, 35 females), ages 16-35, were recruited. These subjects were eligible to participate if they had a body mass index of less than 30 kg/m^2, were experienced in and regularly participated (at least 3x/week) in athletic activities which include jumping, landing, and quick deceleration with change of direction. In order to minimize the potentially confounding effects of testing subjects who use compensatory movement strategies due to pain or dysfunction, subjects were excluded if they had a history of lower extremity orthopedic surgery, injury to the knee ligaments, or if they currently had a lower extremity injury or pain. Additionally, because DXA is contraindicated during pregnancy, females were excluded from participating if they were pregnant.

Instrumentation

Body Composition Testing

Body composition testing was performed in the Nutrition Assessment Research Laboratory on the UNCG campus. Body height and mass were first measured with a digital wall-mounted stadiometer and digital scale. These data were entered into the

EnCORE 2007 DXA software (*GE Healthcare, Madison WI*) and body composition was assessed with the Lunar Prodigy Advance (*GE Healthcare, Madison WI*) fan-beam dual energy x-ray absorptiometry (DXA). EnCORE 2007 software (*GE Healthcare, Madison WI*) was then used to analyze lower extremity lean mass (LELM).

Strength Testing

Maximal isometric, concentric, and eccentric thigh strength was assessed with the Biodex System 3 isokinetic dynamometer (*Biodex Medical Inc., Shirley NY*). Non-normalized peak torques were manually entered into a spreadsheet and averaged across three trials.

Biomechanical Testing

Participants were outfitted with standardized athletic shoes (adidas Uraha 2, *adidas North America, Portland OR*) and then instrumented with 12 active optical LED markers (*Phase Space, San Leandro, CA*) placed on the dominant foot, shank, thigh, and sacrum. The dominant leg was defined as the stance leg when kicking a ball for maximum distance. The markers were affixed and secured to the segments with standard hook and loop material. Kinematics and kinetics were measured with an 8-camera IMPULSE motion tracking system (*Phase Space, San Leandro CA*) at 240 Hz and two non-conducting force platforms (Type 4060-130, *Bertec Corporation, Columbus OH*) at 1000 Hz, respectively, which were interfaced with Motion Monitor software *(Innovative Sports Training, Chicago IL).*

Procedures

Pre-screening and Enrollment of Male-Female Pairs

Individuals who were interested in participating in this study communicated with the investigator via email, phone, or in person to confirm that they met the study requirements. If they qualified for the study and still wished to participate, they were asked to self-report their body height (m) and mass (kg) so that their body mass index (BMI; mass/height^2) could be calculated. If they were unsure of their body mass or height, they visited the lab so that the investigator could take these measurements. Once a male and female with similar BMI's (± 1.0 kg/m^2) were identified, they were enrolled in the study. At that time, participants provided their informed consent according to university IRB protocol.

Test Session One: Body Composition Testing & Calculation of Relative Task Demands

Participants reported to the Nutrition Assessment Research Laboratory on the UNCG campus for body composition assessment with the Lunar Prodigy Advance (*GE Healthcare, Madison WI*) fan-beam dual energy x-ray absorptiometer (DXA). In order to ensure similar hydration levels, all participants received standard instructions to prepare for their scan which required that they: 1) did not exercise or drink alcohol within 24 hours of testing, 2) drank at least 1 liter of water the evening before their scan, 3) did not intake caffeine or food within 2 hours, and 4) voided their bladder within 15 minutes of measurement. Additionally, because DXA is contraindicated during pregnancy, females were required to submit a urine sample so that a pregnancy test could confirm that they were not pregnant before body composition testing could be performed.

While wearing light athletic clothing void of any metal, body height (in) and mass (lb) were measured with a wall-mounted stadiometer and digital scale, respectively. These data were entered into the patient database in the EnCORE 2007 software (*GE Healthcare, Madison WI*) for use in the software's calculations of body composition. The participant was then placed on the DXA table in a supine position, centered on the midline of the table with their arms extended at their sides. In order to ensure the subject's spine was straight, manual traction was applied evenly to both legs until the subject's body slid on the table. Likewise, manual traction was applied to both arms to ensure that the shoulders were at the same height. In order to ensure consistent positioning, the legs were rotated as needed so that both patellae were facing straight up and a strap was applied around the distal tibias to maintain that leg position. Once positioned, the participant was instructed to lie still for the duration of the total body scan, which typically lasted approximately 6 minutes. EnCORE 2007 software (*GE Healthcare, Madison WI*) was then used to quantify Lower Extremity Lean Mass (LELM; kg) (see Data Processing and Reduction).

From these data, the relative task difficulty (lower extremity lean mass to potential energy ratio; $LELM{*}PE^{-1}$) during the drop jump landing was calculated for each participant to represent the amount of available LELM relative to the amount of energy which must be dissipated upon landing (Table 1, column A) from the 0.45m drop height ($LELM{*}PE^{-1}_{STD}$). To achieve this, the relative task difficulty was calculated based on the potential energy (body mass (kg) x gravity (m/s^2) x drop height (m)) of the subject when standing on top of the 0.45m ($Height_{STD}$) box. Based on the difference in relative difficulty between matched female-male pairs, the drop height was increased for the male (Table 1, column B) during the $Height_{EQU}$ condition in order to equalize their task difficulty to their matched female ($LELM*PE^{-1}_{EQU}$; Table 1, column C). For detailed formulas used to calculate the increased drop height, see Appendix B.

The following pilot data are provided to illustrate the method by which relative demands ($LELM*PE^{-1}$) are equalized within a male-female BMI-matched pair

Table 1. Calculation of relative task difficulty ($LELM*PE^{-1}$)for BMI-matched males and females for the $Height_{STD}$ drop height of 0.45 m (column A). Drop height was increased for each male to match his task demand to his female counterpart ($Height_{EQU}$; column B), resulting in equalized relative task demands (column C).

					A	B		C
Sex	**Mass (kg)**	**BMI ($kg*m^{-2}$)**	**LELM (kg)**	**PE_{STD} (J)**	**$LELM*PE^{-1}_{STD}$ ($kg*J^{-1}$)**	**$Height_{EQU}$ (m)**	**PE_{EQU} (J)**	**$LELM*PE^{-1}_{EQU}$ ($kg*J^{-1}$)**
F	62.7	24.10	15.3	276.5	0.055			0.055
M	78.3	24.40	23.1	345.3	0.067	0.54	417.5	0.055

LELM: lower extremity lean mass
$LELM*PE^{-1}_{STD}$: relative difficulty from the standard height of 0.45m.
$Height_{EQU}$: elevated drop height for the male
PE_{EQU}: Potential energy of the male while standing atop the box at the elevated height
$LELM*PE^{-1}_{EQU}$: Relative difficulty for female from $Height_{STD}$ and male from $Height_{EQU}$.

Test Session Two: Familiarization

In order to ensure consistent performance during the strength and biomechanical testing, all participants were extensively familiarized to the testing procedures approximately 7 days prior to collecting strength and biomechanical data.

Dynamic Flexibility Warm-up

In order to prepare the participants for dynamic muscle contractions and to reduce the risk of undue muscle soreness and injury, participants began the familiarization session by performing a standardized dynamic flexibility routine to actively warm and stretch the lower extremity musculature (Table 2). This routine consists of 3-4 minutes of active tissue warming (jogging forwards and backwards and side-shuffling at a self-selected pace), followed by an 8-minute standardized dynamic flexibility warm-up. This protocol consists of various exercises which were performed over a distance of 8 meters, followed by a 10 meter accelerative run, and an 18 meter return jog. See Appendix D for detailed descriptions of each exercise.

Table 2. Dynamic flexibility warm-up protocol performed before familiarization and data collection sessions.

1.	3 minutes of jogging and skipping at self-selected pace
2.	Heel-Toe Walk
3.	Walking Calf Stretch (straight knee)
4.	Easy Alternate Leg Heel Kicks
5.	Progressive Alternate Leg Heel Kicks
6.	Walking Quadriceps Stretch
7.	Backwards Run
8.	Walking Hamstrings
9.	Backwards Run
10.	Walking Toe Touch
11.	Backwards Run
12.	Single Leg Kick
13.	Side Shuffle (alternating- forwards)
14.	Side Shuffle (alternating- backwards)
15.	Walking Side Lunge (alternating legs)
16.	Open the Gate
17.	Close the Gate
18.	Leg Swings

Strength Testing Familiarization

Following the warm-up, participants were familiarized to the strength testing protocol. They were positioned in the Biodex System 3 isokinetic dynamometer (Biodex Medical Systems, Inc., Shirley, NY) using standard procedures: seated in 90° of hip flexion, with the distal portion of the femur lined up with the edge of the seat and the lateral femoral epicondyle aligned with the axis of rotation on the dynamometer. Straps secured the chest, hips, thigh, and distal shank to ensure a constant body position. The participants were then instructed in the strength testing protocol for the quadriceps first, followed by the hamstrings. Subjects were familiarized to each contraction speed for both muscle groups in the same order: isometric, 60º/s, and 180º/s.

Isometric Strength Protocol

For the maximal voluntary isometric contraction (MVIC) strength testing, the quadriceps and hamstrings were tested in a static position of 45 degrees of knee flexion. The participants were instructed to engage the quadriceps by "kicking out against the pad", which remained stationary, and to maintain the contraction for 3 seconds. Once the participant was comfortable with the isometric contractions, they performed submaximal contractions (24%, 50% and 75%) and one maximal (100%) effort practice contractions. This was followed by three maximum effort trials, separated by 30 seconds between trials, during which torque data were recorded. The investigator then assessed the maximal effort trials for consistency and perception of maximal effort in order to determine if the participant was sufficiently familiarized. If the investigator did not believe that the participant was sufficiently familiarized, they were asked to repeat the maximum effort trials and their subsequent performance was re-assessed.

Isokinetic Strength Protocol

To set up range of motion for the isokinetic contractions, the knee was placed in 90 degrees of flexion. Then the knee was extended 70 degrees, which represents the maximum knee extension angle during testing. Hence, isokinetic strength testing occurred throughout a range of motion from 20-90 degrees.

Participants were then instructed on performing the quadriceps testing procedure which consisted of a continuous concentric-eccentric knee extension-flexion protocol at two speeds: 60º/s and 180º/s (Bennett, et al., 2008), with the slower speed performed first. In order to perform the cyclic concentric and eccentric contractions of the quadriceps during knee extension and flexion actions, respectively, the participant was instructed to engage the quadriceps by kicking out against the pad maximally for the entire test, while disregarding the direction of motion. Once the participant was comfortable with the concentric-eccentric actions, they performed 1 submaximal (50%-75%) effort practice set of 5 repetitions. If it appeared that the participant was able to perform the contractions correctly, they were asked to perform 1 set of maximal contractions. The investigator assessed the 5 maximal effort repetitions for consistency and perception of maximal effort in order to determine if the participant was sufficiently familiarized. Specifically, the data was inspected to ensure that the eccentric torques were larger than the concentric torques and that the peak torque values were consistent across all trials. If the investigator did not believe that the participant was sufficiently familiarized, they were provided feedback regarding their performance and were asked to repeat the maximal effort trials. Their subsequent performance was then reassessed.

Once participants were familiarized to the quadriceps testing protocol, they were instructed on performance of the hamstrings testing procedure which consisted of a continuous eccentric-concentric knee extension-flexion protocol at two speeds: 60º/s and 180º/s (Bennett, et al., 2008), using the same range of motion established for the quadriceps. The participant was instructed to engage the hamstrings by pulling back against the pad maximally for the entire test, while disregarding the direction of motion. Once the participant was comfortable with the eccentric-concentric actions, they performed 1 submaximal (50%) effort practice set of 5 repetitions. If they felt comfortable performing the contractions, they then performed one set of 5 maximal contractions during which the data were recorded. The investigator then assessed the level of familiarization of the participant in a manner identical to that of the quadriceps.

Biomechanical Familiarization

Following familiarization to the isokinetic strength testing protocol, participants were familiarized to the biomechanical testing protocol which included a box drop jump landing task from two different heights (0.45 and 0.55 meters), the order of which was counterbalanced between pairs, but identical within each pair (see Appendix C). While atop the box, they were asked to assume an initial position whereby they aligned their 1st MTP joint (i.e. ball of foot) with the edge of the box, and placed their hands at the level of their ears. In order to most effectively instruct the participant, the drop jump task was divided into two components: 1) drop landing and 2) vertical jump. To perform the initial component of the task, they were asked to gradually lean forward through their hips so that they are able to fall straight down off the box without stepping or jumping and land evenly on both feet. They repeated the drop landings until they were comfortable and consistent in their performance. They were then asked to add the second component of the task which included a maximal vertical jump immediately following the initial drop landing. The investigator emphasized the importance of performing a maximal vertical jump each time. To perform the entire drop jump landing task, participants were instructed to assume their initial position, drop straight down off the box, land evenly on both feet, perform a maximal vertical jump, and land once again on both feet. They were asked to perform the drop landing and subsequent jump in one fluid motion (i.e. land, load, jump), rather than two separate motions (i.e. land, pause, load, jump). These instructions ensured that the participants would perform the task in a natural and functional manner. Each participant performed the entire task as many times as needed in order to be comfortable and consistent. The investigator also assessed the participant's performance for consistency and perception of maximal effort.

During the familiarization, all participants also practiced from a height of 0.55 meters to represent the increased height ($Height_{EQU}$) from which they would be dropping during the biomechanical testing session. The true height was later calculated based on the relative difference in LELM between the matched male/female. If the calculated $Height_{EQU}$ was greater than 0.60 meters, the participants were asked to return for another familiarization to the specific height from which they would be landing during testing.

Again, the order of heights was counterbalanced and the procedures for the drop jump landing from both heights were identical.

Test Session Three: Biomechanical and Strength Testing

Approximately 7 days following familiarization, participants reported to the Applied Neuromechanics Research Laboratory for biomechanical and strength testing. Additionally, for descriptive purposes, laxity measurements were performed during this session.

Laxity Measurements

Laxity measurements, including anterior knee laxity, genu recurvatum, and general joint laxity were measured for each participant before any physical activity. Anterior knee laxity was measured with a KT-2000 knee arthrometer (*MedMetric Corp., San Diego CA*). During this measurement, the subject was supine, with their distal femur supported by a thigh bolster and secured with a thigh strap to maintain a neutral thigh rotation position indicated by the patella being lined up in the horizontal plane and a knee flexion angle of 25±5°. The distal tibias were stabilized with the manufacturer-supplied foot cradle to prevent rotation. While applying firm pressure to stabilize the patella, an anteriorly directed load of 30 lbs. was applied and the displacement of the tibia relative to the femur was recorded in millimeters. Three trials were performed and the average displacement was recorded.

Genu recurvatum was measured while the participant was supine, with their distal tibias elevated using a bolster so that the posterior aspect of the knee was clear of the table. The participants were asked to actively contract their quadriceps in order to extend the knee as far as possible. A standard goniometer with arm extensions was used to assess the amount of knee hyperextension during the active contraction by measuring the angle formed between the most prominent aspect of the greater trochanter and midline of the longitudinal axis of the lateral malleolus about the vertex located on the lateral femoral epicondyle. Three measurements were taken and the average recorded.

General Joint Laxity was assessed using a modified Beighton-Horan Mobility index (Shultz, et al., 2007; Shultz, et al., 2010). One point (1= Yes) was awarded for achievement of each of the following measurements, for a total possible score of 9: 5th finger hyperextension >90° (bilateral), Thumb Abduction (ability to touch volar aspect of forearm with thumb; bilateral), Elbow hyperextension >10° (bilateral), Standing active Knee Hyperextension >10° (bilateral), and Trunk Flexion (ability to place palms flat on floor while knees are fully extended).

Biomechanical Testing

Following laxity testing, participants performed the dynamic flexibility warm-up in an identical fashion to that during familiarization. Following the warm-up, participants were instrumented with three optical LED markers on each segment (foot, shank, thigh, and pelvis) for biomechanical analysis. Body mass and height were measured using the force platform and the digitizing stylus, respectively, to enable anthropometric modeling in the Motion Monitor software as well as normalization of kinetic and energetic data. Hip joint centers were calculated using the Leardini method (Leardini et al., 1999), while the knee and ankle joint centers were calculated as the middle of the medial and lateral femoral epicondyles and malleoli, respectively (Madigan & Pidcoe, 2003). The reference system used for kinematic data was established for each segment with the positive Z-axis defined as the left to right axis, the positive Y-axis defined as the distal to proximal vertical axis, and the positive X-axis defined as the posterior-anterior axis. 3-dimensional hip, knee, and ankle flexion angles were calculated using Euler angle definitions with a rotational sequence of ZY'X" (Kadaba et al., 1989).

Participants performed the drop jump landing task during both conditions ($Height_{STD}$, $Height_{EQU}$) according to the appropriate counterbalance order. 5 successful trials at each height were performed from each height. For a trial to be considered successful, the participant had to keep their hands by both ears, drop off the box without stepping or jumping, land evenly on two feet, perform a maximal vertical jump and once again land evenly on two feet. Additionally, the investigator assessed the data collected following each trial to ensure that it was reasonably void of excess noise.

Strength Testing

Following the biomechanical testing protocol, participants performed the maximal strength testing protocols. Consistent with the order of the familiarization, the quadriceps MVIC was tested first. Participants performed three sub-maximal (25%, 50%, and 75%) and one maximal (100%) MVIC trials to re-familiarize themselves to the proper action, followed by 90 seconds rest. Then they performed one set of three maximal effort trials from which the data was extracted. After 90 seconds of rest, they completed five submaximal repetitions at 60°/s, followed by 90 seconds of rest, and five maximal repetitions at this speed. They then repeated this procedure at the 180°/s speed. Following quadriceps testing, participants performed the hamstring testing in an identical fashion.

Data Processing and Reduction

Body Composition Data

ENCORE 2007 software (*GE Healthcare, Madison WI*) was used to partition total body composition into regional bone, lean, and fat mass so that relative amounts and locations of lean mass can be determined. The standard regions of interest (ROI) provided within the software were manually adjusted using defined anatomical landmarks and boundaries. Specifically, the lower extremity ROI was formed by a diagonal line which extended from the iliac crest through the femoral neck and then continued down to the tip of the longest toe, and laterally to include all soft tissue of the leg (see Figure 1). For additional ROI definitions, see Appendix A. The principle investigator has established excellent day- to-day reliability ($ICC_{2,1}$=0.94-1.00) for manual lower extremity ROI placement in a group of 15 recreationally-active females (mean age= 20.3±2.4 yrs, BMI= 23.2±3.4 kg/m^2).

Figure 1. Standard regions of interest (ROI) are manually-adjusted to defined the lower extremity.

Strength Data

Gravity-corrected torque data from the Biodex System 3 software were exported. Custom software was used to extract, filter (10Hz), and window (cut off torque occurring <70% of dynamometer speed) the peak non-normalized torques (Nm) and work values (J) for the quadriceps and hamstrings at each speed. All data were visually inspected for noise (i.e. artificial peaks) before being imported into a spreadsheet in Microsoft Excel 2007 (*Microsoft Corp, Redmond WA*). For the isometric contractions, the average peak torque from all 3 trials (<8.0% CV) were used. For the isokinetic contractions, each contraction was visually inspected to ensure that no excess noise was present. Then the average peak torques across the 3 highest contractions (< 8% CV) were used for analysis. For the purposes of this study, the eccentric torque data were used.

Biomechanical Data

A 4^{th} order, zero-lag low pass Butterworth filter was used to process the kinematic (12Hz) and kinetic (12Hz) data. Processing was performed with Motion Monitor software (*InnSports Training, Chicago IL).* All data were then exported to Microsoft Excel 2007 (*Microsoft Corp, Redmond WA*) for further reduction. From these data, custom software extracted hip, knee, and ankle joint excursion from initial contact (vGRF>10N) to peak center of mass displacement and was calculated as the average value obtained across the five trials. Hip, knee, and ankle moments were calculated using inverse dynamics solutions. Energy absorption (eccentric work of hip extensors, knee extensors, and ankle extensors) was calculated as the area under the negative power curve (Zhang, et al., 2000; Decker, et al., 2003; Schmitz & Shultz, 2010) using the following formulas:

- Power (J)= Moment (Nm) x Angular Velocity (rad/s)
- Moment (Nm): calculated from inverse dynamics: (Force (N) x Moment arm (m))
- Angular Velocity (rad*s^{-1}): 1st derivative of position at each time interval (Position x Time)

For all analyses, joint moments and energy absorption were normalized to body weight (N)*height (m) while LELM and strength variables were normalized to body mass (kg).

Data Analysis

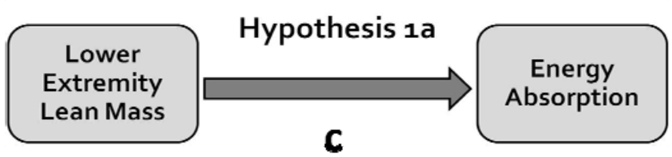
Statistical Plan

The following statistical plan was used to test each hypothesis:

1. To determine whether less lower extremity lean mass predicted less energy absorption (Hypothesis 1a), and whether this relationship was mediated by eccentric thigh strength (Hypothesis 1b), a mediation analysis was performed (Baron & Kenny, 1986; Preacher & Hayes, 2004).

This was achieved by first examining the strength of the relationship between LELM and EA during the $Height_{STD}$ condition (Figure 2) with a linear regression model (Baron & Kenny, 1986) which determined the "total effect" of LELM on EA (Preacher & Hayes, 2004).

Figure 2: Total effect of lower extremity lean mass on energy absorption: the extent to which lower extremity lean mass (LELM) predicts energy absorption (EA)

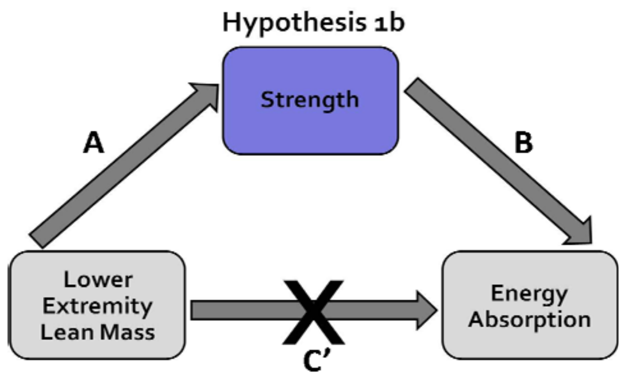


$$\text{Energy Absorption} = \beta_0 + \beta_1 \text{ (LELM)} + e$$

Next, the individual relationships between LELM and strength (Fig. 3, Path A) and strength and energy absorption (Fig. 3, Path B) (Baron & Kenny, 1986) were examined. Finally, the "direct relationship" (Preacher & Hayes, 2004) between LELM and energy absorption after controlling for the proposed mediator, strength (Fig. 3, Path

C'), was examined. This variable is said to be a mediator if the relationship between LELM and energy absorption is decreased after including strength in the model (Baron & Kenny, 1986).

Figure 3. Direct relationship between lower extremity lean mass and energy absorption, mediated by strength



$$\text{Energy Absorption} = \beta_0 + \beta_1\,(\text{LELM}) + \beta_2\,(\text{Quad}_{\text{ECC}}) + e$$

$$\text{Energy Absorption} = \beta_0 + \beta_1\,(\text{LELM}) + \beta_2\,(\text{Ham}_{\text{ECC}}) + e$$

Formal tests for significance of the indirect path between LELM and EA, through strength (Quad_{ECC} and Ham_{ECC} at 180°/s) (Path A*Path B, Figure 3) were performed using a Sobel Test and a non-parametric bootstrapping procedure (Preacher & Hayes, 2004). The Sobel Test tests whether the strength of the indirect path (Path A*Path B) is not equal to zero (Baron & Kenny, 1986). Significance of the indirect path indicates that the significant relationship between LELM and EA exists only due to the relationship between strength and EA (Preacher & Hayes, 2004; Howell, 2006). The bootstrapping procedure resamples the data with replacement in order to eliminate the assumption of normality of data which is present using the Sobel Test (Preacher & Hayes, 2004). These analyses were ultimately used to test whether the relationship between LELM and EA (Hypothesis 1a) was significantly diminished when $Quad_{ECC}$ or Ham_{ECC} were added to the regression model constructed for Hypothesis 1a.

For Hypothesis 1a, separate tests examined the relationships between LELM and total energy absorption (EA_{TOT}) as well as energy absorption at the hip (EA_{HIP}), knee (EA_{KNEE}), and ankle (EA_{ANK}); these analyses were performed with all subjects, and then stratified by sex. Additionally, for the mediation analysis (Hypothesis 1b), the eccentric strength variables ($Quad_{ECC}$ and Ham_{ECC}) were added in separate models.

2. To determine whether the relationship between eccentric thigh torque and energy absorption was stronger in females vs. males (Hypothesis 2), a stepwise linear multiple regression (probability of removal p= 0.51) was developed for each joint at both heights ($Height_{STD}$ and $Height_{EQU}$).

$$\text{Energy Absorption} = \beta_0 + \beta_1 (\text{Sex}) + [\beta_2 + \beta_3 (\text{Sex})] * \text{Quad}_{ECC} + [\beta_4 + \beta_5 (\text{Sex})] * \text{Ham}_{ECC} + e$$

- Outcome Variables: Energy Absorption (EA_{TOT}, EA_{HIP}, EA_{KNEE}, and EA_{ANK})
- Predictor Variables: Sex, $Quad_{ECC}$ (β_2), Ham_{ECC} (β_4), Sex*$Quad_{ECC}$ (β_3), and Sex*Ham_{ECC} (β_5).

Significant interactions (sex*$Quad_{ECC}$ or sex*Ham_{ECC}) would indicate that the relationships between strength and energy absorption are different across males and females.

3. To determine whether increasing the relative task demand resulted in greater energy absorption (Hypothesis 3), a multivariate repeated measures design examined normalized hip, knee, and ankle energy absorption ($J*BW^{-1}*m^{-1}$) between the standardized and equalized drop jump landing heights, where the two landing conditions were the repeated measures. If the overall multivariate test was significant, univariate ANOVAs were examined for each joint.

- Multivariate Dependent Variables: Energy absorption (EA_{HIP}, EA_{KNEE}, EA_{ANK})
- Within factor: Repeated Conditions ($Height_{STD}$, $Height_{EQU}$)
- Between factor: Sex (male, female)

Further, to examine whether equalizing the relative task demands (by comparing $Height_{STD}$ in females vs. $Height_{EQU}$ in males) results in similar energy absorption strategies between size-matched males and females (Hypothesis 3b), separate one-way ANOVAs compared males and females on EA_{TOT}, EA_{HIP}, EA_{KNEE}, EA_{ANK}.

Lastly, to determine whether increasing task demands resulted in greater changes in energy absorption strategies in females compared to males from the $Height_{STD}$ to the $Height_{EQU}$ because of the greater overall demands of females vs. males (Hypothesis 3c), the interaction term from the multivariate model was examined by comparing males and females on the delta change between $Height_{STD}$ and $Height_{EQU}$.

Power Analysis

Power Analyses (G*Power 3) were performed *a priori* for each statistical model to calculate the sample sizes needed to achieve 80% power at a significance level of α=0.05:

For a linear multiple regression model with 1 predictor (Hypothesis 1a), 25 subjects per group (males and females) were needed to detect an r^2= 0.26 (with a moderate effect size =0.35).

For a linear multiple regression model with 2 predictors (Hypothesis 1b), 31 subjects per group (males and females) were needed to detect an r^2= 0.26 (with a moderate effect size =0.35).

For a linear multiple regression model with 5 predictors (Hypothesis 2), a total sample size of 43 subjects were needed to detect an r^2= 0.26 (with a moderate effect size =0.35).

For a multivariate repeated measures design with 2 groups and 2 conditions (Hypotheses 3a, 3c) with a moderate effect size of 0.35, a total sample of 67 (34 per group) is needed to detect significant differences.

For a one-way ANOVA with 2 groups (Hypothesis 3b) with a moderate effect size of 0.35, a total sample of 68 (34 per group) is needed to detect significant differences.

CHAPTER IV

RESULTS

Descriptives

35 males and 35 females successfully completed data collection. Table 1 lists the demographic and body composition characteristics of all subjects and when stratified by sex. As expected, males were taller (1.78±0.06 vs. 1.67±0.06 m., $p<0.001$) and heavier (74.7±8.9 vs. 65.3±6.6 kg., $p<0.001$) than their female counterparts, but there were no differences in body mass index (BMI) (23.6±2.2 vs. 23.3±2.2 kg/m^2, $p=0.596$). Males also possessed more total body lean mass (61.6±6.5 vs. 43.8±5.8 kg., $p<0.001$) as well as lower extremity lean mass (LELM) in both absolute quantity (21.6±2.8 vs. 15.2±1.9 kg., $p<0.001$) and as a proportion of total body mass (29.1±2.3 vs. 23.6±2.9%, $p<0.001$) compared to females. However, when LELM was expressed as a proportion of total body lean mass, there was no sex difference in relative lean mass distribution in the lower extremity (35.2±1.7 vs. 35.1±1.7 %, $p=0.887$).

Table 3. Demographic and body composition descriptives for all subjects (n=70), and also stratified by sex. All values are expressed as mean±SD and range (min-max).

	ALL SUBJECTS (n=70)	**FEMALES** (n=35)	**MALES** (n=35)	**p-value**	**Effect Size**
Age (yrs)	21.3±3.3 (17.0-32.0)	21.6±3.6 (17.0-32.0)	20.9±2.9 (18.0-32.0)	p=0.362	0.25
Height (m)	1.72±0.1 (1.53-1.87)	1.67±0.1 (1.50-1.80)	1.78±0.1* (1.66-1.87)	p<0.001	1.05
Mass (kg)	70.0±9.1 (48.9-100.7)	65.3±6.6 (48.9-77.2)	74.7±8.9* (57.3-100.7)	p<0.001	1.86
BMI (kg/m^2)	23.5±2.2 (19.0-29.2)	23.3±2.2 (19.0-29.2)	23.6±2.2 (19.7-29.1)	p=0.596	0.13
TBLM (kg)	52.7±10.9 (32.8-75.3)	43.8±5.8 (32.8-67.0)	61.6±6.5* (49.6-75.3)	p<0.001	2.74
LELM (kg)	18.4±4.0 (11.3-27.3)	15.2±1.9 (11.3-19.8)	21.6±2.8* (16.9-27.3)	p<0.001	2.31
LELM (% Body Mass)	26.3±3.8 (19.0-35.0)	23.6±2.9 (19.0-32.0)	29.1±2.3* (24.0-35.0)	p<0.001	2.34
LELM (% Lean Mass)	35.1±1.7 (31.0-40.0)	35.1±1.7 (31.0-40.0)	35.2±1.7 (33.0-39.0)	p=0.887	0.04

* Males >Females, p<0.05
BMI: Body mass index
TBLM: Total body lean mass
LELM: Lower extremity lean mass

Table 4 presents the strength testing results for the specific variables of interest: eccentric quadriceps and hamstring contractions at 180°/s ($Quad_{ECC}$ and Ham_{ECC}) for all subjects, and when stratified by sex. Specific to $Quad_{ECC}$ and Ham_{ECC}, males produced 21.9% and 25.0% larger peak torques, respectively, compared to females. However, when normalizing to LELM of the test leg, males and females had similar $Quad_{ECC}$ and Ham_{ECC} values (Appendix E).

Table 4. Strength testing results for all subjects and when stratified by sex. Mean±SD, range (min-max), and between-subject effect size are provided for eccentric quadriceps ($Quad_{ECC}$) and hamstring (Ham_{ECC}) peak torque values.

	ALL SUBJECTS (n=70)	**FEMALES (n=35)**	**MALES (n=35)**	**p-value**	**Effect Size**
$Quad_{ECC}{}^{a}$	3.5±0.7 (1.8-5.7)	3.2±0.5 (1.8-4.0)	3.9±0.7* (2.0-5.7)	p<0.001	0.99
$Ham_{ECC}{}^{a}$	2.2±0.4 (1.3-3.1)	2.0±0.3 (1.3-2.5)	2.5±0.3* (2.0-3.1)	p<0.001	2.07
$Quad_{ECC}{}^{b}$	27.1±4.8 (14.3-40.4)	27.2±4.7 (18.2-40.4)	27.0±4.9 (14.3-35.4)	p=0.824	0.05
$Ham_{ECC}{}^{b}$	16.9±2.1 (11.1-21.6)	16.5±2.1 (11.1-20.6)	17.2±2.1 (12.4-21.6)	p=0.187	0.32

* Indicates Males > Females (p<0.05)

[a] Normalized to Total Body Mass (Nm/kg)

[b] Normalized to Lower Extremity Lean Mass (Nm/kg)

Note: values for isometric and concentric contractions are provided in Appendix E.

Table 5 presents the mean±SD and between-subject effect sizes for the energy absorption data during each condition ($Height_{STD}$ and $Height_{EQU}$) for all subjects and when stratified by sex. To aid in interpretation of the energetics data (primary outcome variables), kinetic and kinematic data are presented in Table 6.

Table 5. Energy absorption results (mean±SD) for both conditions separately ($Height_{STD}$ and $Height_{EQU}$) and when task difficulty was equalized ($Height_{EQU}$ in males vs. $Height_{STD}$ in females). Effect sizes are provided for sex and condition.

ENERGY ABSORPTION (J x N^{-1} x m^{-1})†					
		All (n=70)	**Females (n=35)**	**Males (n=35)**	**Between-Sex Effect Size**
$Height_{STD}$	Total	16.2±2.9	15.0±2.7	17.3±2.6	0.88
(0.45±0.00 meters)	Hip	3.3±1.6	2.7±1.3	3.9±1.7	0.68
	Knee	9.4±2.6	9.1±2.6	9.8±2.6	0.24
	Ankle	3.4±1.4	3.1±1.4	3.7±1.2	0.44
$Height_{EQU}$	Total	17.5±3.2	16.2±2.8	18.8±3.2	0.80
(0.57±0.07 meters)	Hip	3.7±1.7	3.0±1.4	4.3±1.8	0.70
	Knee	10.3±2.9	10.0±2.5	10.6±3.2	0.20
	Ankle	3.5±1.5	3.2±1.5	3.8±1.4	0.45
Between-Condition Effect Size	Total	0.41	0.44	0.45	
	Hip	0.22	0.24	0.24	
	Knee	0.29	0.33	0.26	
	Ankle	0.08	0.05	0.11	
Equalized Condition	Total		15.0±2.7	18.8±3.2**	1.18
	Hip		2.7±1.3	4.3±1.8**	0.88
	Knee		9.1±2.6	10.6±3.2*	0.46
	Ankle		3.1±1.4	3.8±1.4*	0.50

† Values are expressed x10^2
* Males > Females ($p<0.05$)
** Males > Females ($p<0.001$)

Table 6. Kinetic and kinematic results for all subjects (n=70) and when stratified by sex (n=35) during $Height_{STD}$ (0.45±0.00 m) and $Height_{EQU}$ (0.57±0.07 m) conditions. Mean±SD and between-subject effect sizes are presented for each variable.

PEAK EXTENSOR MOMENTS ($Nm*N^{-1}*m^{-1}$)					
		ALL	**FEMALES**	**MALES**	**Effect Size**
$Height_{STD}$	Hip	1.22±0.33	1.09±0.25	1.36±0.35	0.78
	Knee	1.12±0.25	1.07±0.26	1.17±0.24	0.42
	Ankle	0.65±0.16	0.59±0.14	0.71±0.16	0.70
$Height_{EQU}$	Hip	1.35±0.35	1.24±0.26	1.46±0.39	0.57
	Knee	1.14±0.26	1.10±0.23	1.19±0.28	0.32
	Ankle	0.65±0.16	0.59±0.13	0.71±0.17	0.68
INITIAL FLEXION ANGLE (Deg)					
		ALL	**FEMALES**	**MALES**	**Effect Size**
$Height_{STD}$	Hip	11.6±7.5	12.1±7.7	11.2±7.4	0.13
	Knee	14.0±8.1	11.3±7.6	16.8±7.6	0.73
	Ankle	51.7±9.2	49.5±9.0	53.8±9.0	0.48
$Height_{EQU}$	Hip	10.2±6.6	10.9±6.6	9.5±6.7	0.20
	Knee	12.4±7.1	9.9±6.8	14.9±6.7	0.76
	Ankle	49.7±8.2	47.8±8.6	51.6±7.4	0.52
TOTAL JOINT EXCURSION (Deg)					
		ALL	**FEMALES**	**MALES**	**Effect Size**
$Height_{STD}$	Hip	51.1±11.4	51.3±12.4	51.0±10.5	0.03
	Knee	75.2±10.1	73.3±10.6	77.1±9.4	0.40
	Ankle	57.5±8.6	58.5±7.8	56.5±9.3	0.21
$Height_{EQU}$	Hip	53.4±11.9	53.5±12.5	53.3±11.5	0.01
	Knee	77.5±10.1	75.6±10.6	79.4±9.5	0.40
	Ankle	58.8±7.7	59.7±7.3	57.9±8.1	0.22
PEAK VERTICAL GROUND REACTION FORCE (N)					
		ALL	**FEMALES**	**MALES**	**Effect Size**
$Height_{STD}$	vGRF	1.48±0.17	1.50±0.15	1.45±0.19	0.26
$Height_{EQU}$	vGRF	1.68±0.25	1.73±0.23	1.62±0.27	0.44

HYPOTHESIS 1: Relationships between Lean Mass, Strength, and Energy Absorption

Table 7 presents the Pearson correlation coefficients for LELM, $Quad_{ECC}$, Ham_{ECC}, and EA_{TOT}, EA_{HIP}, EA_{KNEE}, and EA_{ANK}.

Table 7. Pearson product correlations between lower extremity lean mass (LELM) [% body mass (kg)], eccentric strength [$Quad_{ECC}$, Ham_{ECC}; (Nm/kg)] and energy absorption (EA_{TOT}, EA_{HIP}, EA_{KNEE} and EA_{ANK}) for all subjects and when stratified by sex.

	All Subjects (n=70)				
	LELM	**Total**	**Hip**	**Knee**	**Ankle**
LELM		0.413*	0.327*	0.248*	0.022
$Quad_{ECC}$	0.595*	0.429*	0.252*	0.298*	0.041
Ham_{ECC}	0.690*	0.405*	0.468*	0.123	0.078
	Females (n=35)				
	LELM	**Total**	**Hip**	**Knee**	**Ankle**
LELM		0.312	-0.052	0.402*	-0.087
$Quad_{ECC}$	0.511*	0.426*	0.008	0.389*	0.088
Ham_{ECC}	0.573*	0.310	0.478*	0.182	-0.174
	Males (n=35)				
	LELM	**Total**	**Hip**	**Knee**	**Ankle**
LELM		0.041	0.244	0.042	-0.332
$Quad_{ECC}$	0.308	0.193	0.132	0.203	-0.208
Ham_{ECC}	0.077	0.045	0.205	-0.080	0.000

* Significant correlation ($p<.05$)

Hypothesis 1a: Relationship between LELM and EA

Tables 8-10 present the beta coefficients and model R^2 for each regression analysis describing the extent to which LELM predicted energy absorption for all subjects, females, and males, respectively. LELM was a positive predictor of EA_{TOT} (R^2=

0.159, p<0.001), EA_{HIP}, (R^2=0.094, p=0.006) and EA_{KNEE} (R^2=0.048, p=0.038) when all subjects were analyzed together (Table 7). Within females, LELM was a significant predictor of EA_{KNEE} (R^2= 0.136, p=0.017) (Table 8), but not in males (Table 9). No other models were significant.

Table 8. Regression coefficients and model R^2 when predicting energy absorption (E_{ATOT}, EA_{HIP}, EA_{KNEE}, and EA_{ANK}) with lower extremity lean mass (LELM) [% body mass (kg)] for all subjects (n=70).

Variable	Unstandardized Beta	Standard Error	Standardized Beta	t-value	Adjusted R^2	p-value
Total Energy Absorption ($J*N^{-1}*m^{-1}$)						
Intercept	0.078	0.023		3.474		
LELM	0.317	0.085	0.413*	3.743	0.159*	0.000
Hip Energy Absorption ($J*N^{-1}*m^{-1}$)						
Intercept	-0.004	0.013		-0.294		
LELM	0.139	0.049	0.327*	2.849	0.094*	0.006
Knee Energy Absorption ($J*N^{-1}*m^{-1}$)						
Intercept	0.049	0.022		2.270		
LELM	0.172	0.081	0.248*	2.112	0.048*	0.038
Ankle Energy Absorption ($J*N^{-1}*m^{-1}$)						
Intercept	0.032	0.012		2.774		
LELM	0.008	0.044	0.022	0.181	-0.014	0.857

* Significant coefficient or model (p<.05)

Table 9. Regression coefficients and model R^2 when predicting energy absorption (E_{ATOT}, EA_{HIP}, EA_{KNEE}, and EA_{ANK}) with lower extremity lean mass (LELM) [% body mass (kg)] for females (n=35).

Variable	Unstandardized Beta	Standard Error	Standardized Beta	t-value	Adjusted R^2	p-value
Total Energy Absorption ($J*N^{-1}*m^{-1}$)						
Intercept	0.081	0.037		2.175		
LELM	0.294	0.156	0.312	1.884	0.070	0.068
Hip Energy Absorption ($J*N^{-1}*m^{-1}$)						
Intercept	0.033	0.019		1.750		
LELM	-0.023	0.078	-0.052	-0.298	-0.028	0.768
Knee Energy Absorption ($J*N^{-1}*m^{-1}$)						
Intercept	0.005	0.034		0.157		
LELM	0.364	0.144	0.402*	2.520	0.136*	0.017
Ankle Energy Absorption ($J*N^{-1}*m^{-1}$)						
Intercept	0.042	0.021		2.022	-0.023	0.621
LELM	-0.043	0.087	-0.087	-0.500		

* Significant coefficient or model ($p<.05$)

Table 10. Regression coefficients and model R^2 when predicting energy absorption (E_{ATOT}, EA_{HIP}, EA_{KNEE}, and EA_{ANK}) with lower extremity lean mass (LELM) [% body mass (kg)] for males (n=35).

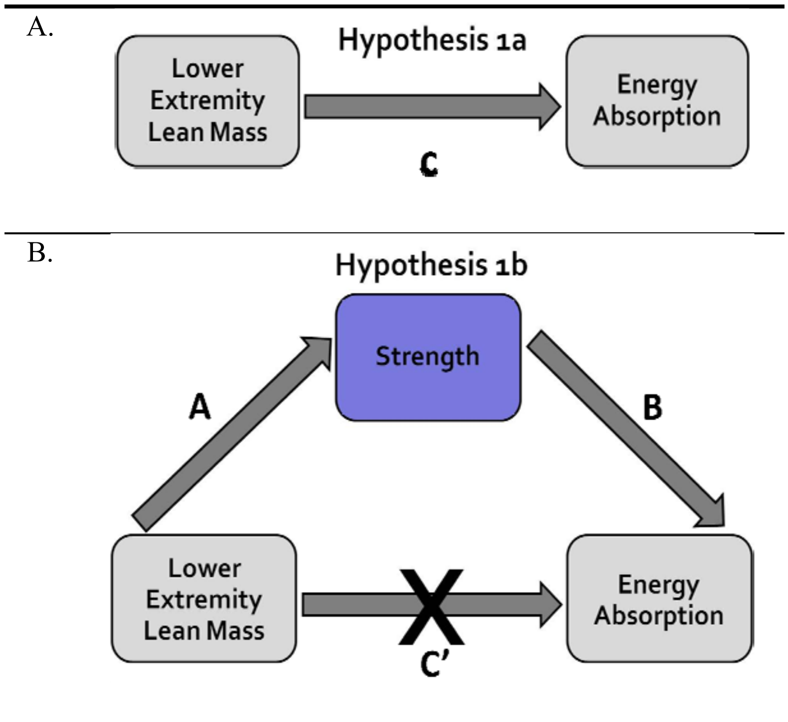
Variable	Unstandardized Beta	Standard Error	Standardized Beta	t-value	Adjusted R^2	p-value
Total Energy Absorption ($J*N^{-1}*m^{-1}$)						
Intercept	0.160	0.057		2.782		
LELM	0.046	0.197	0.041	0.234	-0.029	0.816
Hip Energy Absorption ($J*N^{-1}*m^{-1}$)						
Intercept	-0.013	0.036		-0.369		
LELM	0.179	0.124	0.244	1.444	0.031	0.158
Knee Energy Absorption ($J*N^{-1}*m^{-1}$)						
Intercept	0.084	0.058		1.454		
LELM	0.048	0.197	0.042	0.242	-0.028	0.810
Ankle Energy Absorption ($J*N^{-1}*m^{-1}$)						
Intercept	0.088	0.025		3.468		
LELM	-0.177	0.087	-0.332	-2.025	0.068	0.051

* Significant coefficient or model ($p<.05$)

Hypothesis 1b: Strength as a Mediator of the Relationship between LELM and EA

Figure 4 illustrates the mediation model used to determine whether $Quad_{ECC}$ and Ham_{ECC} mediated the relationship between LELM and EA. Using $Quad_{ECC}$ as an example, Path C represents the direct relationship between LELM and EA (Figure 4, panel A). Path A represents the extent to which LELM predicts $Quad_{ECC}$, while Path B represents the relationship between $Quad_{ECC}$ and EA. Finally, C' demonstrates the extent to which LELM predicts EA, after controlling for $Quad_{ECC}$ (Figure 4, panel B).

Figure 4. Mediation pathways for Hypothesis 1.



According to the criteria for simple mediation as described by Baron and Kenny (Baron & Kenny, 1986), a variable may be considered a mediator if Paths C, A, and B are significant along with a subsequent non-significant relationship between LELM and EA (Path C') once the hypothesized mediator ($Quad_{ECC}$ or Ham_{ECC}) is included in the model. However, as current practice in mediation analysis dictates (Shrout & Bolger, 2002; Hayes, 2009), two methods which formally test the significance of the indirect path (Sobel Test and bootstrapping procedure) were ultimately used in this analysis. The Sobel Test, which is expressed as the product of Path A and Path B (Path A * Path B), describes the difference in relationship between LELM and EA before and after including $Quad_{ECC}$ or Ham_{ECC} in the model. The bootstrapping procedure provides an inference about the size of the indirect effect (mean of Path A * Path B after 5000 resamples), a standard error estimate, and 95% confidence interval for the distribution of the 5000 resamples. Results for these tests indicate whether there is a difference in the relationship between LELM after including eccentric strength ($Quad_{ECC}$ or Ham_{ECC}) with the null hypothesis being that there is no difference between the two models. Separate analyses were performed for $Quad_{ECC}$ and Ham_{ECC} and for all EA variables with all subjects together, and then stratified by sex. Tables 11 and 12 display the results of these two formal tests. Individual path results for each simple mediation analysis, predicting EA with LELM and $Quad_{ECC}$ and Ham_{ECC} are provided in Appendix G.

Table 11. Results for direct tests of significance of indirect paths for $Quad_{ECC}$ as a mediator of the relationship between LELM and EA_{TOT}, EA_{HIP}, EA_{KNEE}, and EA_{ANK}. Separate analyses were performed for all subjects and when stratified by sex.

ALL	Value	SE	95% CI		Z-score	Sig.
Total Energy Absorption						
Sobel	0.130	0.066	0.001	0.258	1.979	0.04*
Bootstrap (5000 resamples)	0.132	0.070	0.011	0.293*		
Hip Energy Absorption						
Sobel	0.023	0.037	-0.050	0.095	0.608	0.543
Bootstrap (5000 resamples)	0.021	0.045	-0.068	0.111		
Knee Energy Absorption						
Sobel	0.096	0.062	-0.026	0.219	1.543	0.123
Bootstrap (5000 resamples)	0.098	0.070	-0.032	0.252		
Ankle Energy Absorption						
Sobel	0.009	0.033	-0.055	0.074	0.278	0.781
Bootstrap (5000 resamples)	0.010	0.034	-0.054	0.081		
FEMALES	**Value**	**SE**	**95% CI**		**Z-score**	**Sig.**
Total Energy Absorption						
Sobel	0.174	0.106	-0.034	0.381	1.644	0.100
Bootstrap (5000 resamples)	0.177	0.107	0.008	0.423*		
Hip Energy Absorption						
Sobel	0.011	0.049	-0.086	0.108	0.218	0.827
Bootstrap (5000 resamples)	0.007	0.052	-0.107	0.109		
Knee Energy Absorption						
Sobel	0.115	0.095	-0.070	0.301	1.219	0.223
Bootstrap (5000 resamples)	0.118	0.091	-0.036	0.330		
Ankle Energy Absorption						
Sobel	0.046	0.056	-0.063	0.155	0.822	0.411
Bootstrap (5000 resamples)	0.051	0.055	-0.029	0.187		
MALES	**Value**	**SE**	**95% CI**		**Z-score**	**Sig.**
Total Energy Absorption						
Sobel	0.069	0.081	-0.090	0.229	0.854	0.393
Bootstrap (5000 resamples)	0.082	0.095	-0.055	0.325		
Hip Energy Absorption						
Sobel	0.014	0.047	-0.077	0.106	0.302	0.763
Bootstrap (5000 resamples)	0.018	0.058	-0.086	0.153		
Knee Energy Absorption						
Sobel	0.074	0.082	-0.088	0.235	0.894	0.371
Bootstrap (5000 resamples)	0.084	0.097	-0.065	0.323		
Ankle Energy Absorption						
Sobel	-0.019	0.034	-0.086	0.048	-0.564	0.573
Bootstrap (5000 resamples)	-0.023	0.040	-0.118	0.044		

* Significant mediation effect ($p<0.05$)

Table 12. Results for direct tests of significance of indirect paths for Ham_{ECC} as a mediator of the relationship between LELM and EA_{TOT}, EA_{HIP}, EA_{KNEE}, and EA_{ANK}. Separate analyses were performed for all subjects and when stratified by sex.

ALL	Value	SE	95% CI		Z-score	Sig.
Total Energy Absorption						
Sobel	0.121	0.082	-0.040	0.282	1.478	0.139
Bootstrap (5000 resamples)	0.127	0.081	-0.018	0.299		
Hip Energy Absorption						
Sobel	0.137	0.048	0.044	0.230	2.874	0.004*
Bootstrap (5000 resamples)	0.140	0.052	0.050	0.255*		
Knee Energy Absorption						
Sobel	-0.044	0.079	-0.199	0.111	-0.557	0.578
Bootstrap (5000 resamples)	-0.038	0.084	-0.203	0.134		
Ankle Energy Absorption						
Sobel	0.030	0.042	-0.053	0.113	0.708	0.479
Bootstrap (5000 resamples)	0.027	0.039	-0.051	0.101		
FEMALES	**Value**	**SE**	**95% CI**		**Z-score**	**Sig.**
Total Energy Absorption						
Sobel	0.106	0.115	-0.120	0.332	0.918	0.359
Bootstrap (5000 resamples)	0.108	0.131	-0.136	0.397		
Hip Energy Absorption						
Sobel	0.195	0.066	0.065	0.325	2.947	0.003*
Bootstrap (5000 resamples)	0.208	0.085	0.076	0.405*		
Knee Energy Absorption						
Sobel	-0.037	0.106	-0.244	0.170	-0.352	0.725
Bootstrap (5000 resamples)	-0.047	0.127	-0.326	0.177		
Ankle Energy Absorption						
Sobel	-0.053	0.064	-0.178	0.073	-0.826	0.409
Bootstrap (5000 resamples)	-0.052	0.054	-0.169	0.047		
MALES	**Value**	**SE**	**95% CI**		**Z-score**	**Sig.**
Total Energy Absorption						
Sobel	0.004	0.039	-0.073	0.080	0.095	0.925
Bootstrap (5000 resamples)	0.011	0.046	-0.072	0.125		
Hip Energy Absorption						
Sobel	0.011	0.034	-0.055	0.076	0.316	0.752
Bootstrap (5000 resamples)	0.012	0.037	-0.052	0.103		
Knee Energy Absorption						
Sobel	-0.007	0.042	-0.089	0.074	-0.177	0.859
Bootstrap (5000 resamples)	0.003	0.051	-0.104	0.120		
Ankle Energy Absorption						
Sobel	0.001	0.017	-0.032	0.035	0.062	0.951
Bootstrap (5000 resamples)	-0.003	0.021	-0.052	0.036		

* Significant mediation effect ($p<0.05$)

When examining results from the Sobel Test, $Quad_{ECC}$ was found to be a significant mediator for EA_{TOT} only when all subjects were analyzed together ($p=0.04$). $Quad_{ECC}$ was not a significant mediator for any EA variables in females or males when analyzed separately. Results following the bootstrapping procedure confirmed the Sobel Test showing that $Quad_{ECC}$ was a significant mediator of the effect of LELM on EA_{TOT} when all subjects were analyzed together (ab= 0.132, 95% C.I. = [0.01, 0.29]), but also when females were analyzed separately (ab= 0.177, 95% C.I. = [0.01, 0.42]). $Quad_{ECC}$ was not a significant mediator for EA_{TOT}, EA_{HIP}, EA_{KNEE}, or EA_{ANK} in males.

Results from both significance tests indicated that Ham_{ECC} was a significant mediator of the relationship between LELM and EA_{HIP} when all subjects were analyzed together ($p=0.004$; ab= 0.14, 95% C.I. = [0.05, 0.26]) and separately in females ($p=0.003$; ab= 0.208, 95% C.I. = [0.07, 0.41]). No mediation effect was present for EA_{TOT}, EA_{KNEE}, or EA_{ANK}. Like $Quad_{ECC}$, Ham_{ECC} was not a significant mediator in any of the models when males were analyzed separately.

Summary of Results for Hypothesis 1

When males and females were analyzed collectively, weak to moderate relationships were observed between LELM and EA_{TOT}, EA_{HIP}, and EA_{KNEE}, suggesting that greater LELM is related to greater EA. When separated by sex, this positive relationship between LELM and EA was only observed for EA_{KNEE} in females and no relationships were observed in males. Similarly, moderate relationships were observed between LELM and $Quad_{ECC}$ and Ham_{ECC} for all subjects together and when females were analyzed separately. There were no relationships between LELM and $Quad_{ECC}$ or Ham_{ECC} in males.

The mediation analyses support a mediating effect of $Quad_{ECC}$ on the relationship between LELM and EA_{TOT}, and Ham_{ECC} on relationship between LELM and EA_{HIP} when males and females are analyzed together. In females only, $Quad_{ECC}$ appeared to mediate the relationship between LELM and EA_{TOT}, while Ham_{ECC} was a mediator between LELM and EA_{HIP}. These findings indicate that the relationships observed between LELM and EA_{TOT} and EA_{HIP} existed primarily through their individual relationships with $Quad_{ECC}$ and Ham_{ECC}, respectively.

Hypothesis 2: Influence of Task Difficulty on Relationships between Strength and Energy Absorption

To determine whether the relationships between strength and energy absorption are influenced by relative task difficulty, separate multiple linear regressions examined the extent to which sex, $Quad_{ECC}$ and Ham_{ECC} predicted EA_{TOT}, EA_{HIP}, EA_{KNEE}, and EA_{ANK} at each height ($Height_{STD}$, $Height_{EQU}$). Sex*$Quad_{ECC}$ and sex*Ham_{ECC} interaction terms were also examined to determine whether the relationship between eccentric strength and energy absorption was stronger for females compared to males due to the greater relative task difficulty for females. Descriptives for these variables were previously presented in Tables 3-5.

Table 13 lists Pearson product correlations between sex, strength, and energy absorption during both landing conditions ($Height_{STD}$ and $Height_{EQU}$).

Table 13. Correlations between lean mass, strength, and energy absorption during both landing conditions. Results are presented for all subjects and when stratified by sex.

		HEIGHT$_{STD}$ †				HEIGHT$_{EQU}$ #			
					All Subjects (n=70)				
	Sex	**Total**	**Hip**	**Knee**	**Ankle**	**Total**	**Hip**	**Knee**	**Ankle**
Sex		0.404*	0.361*	0.120	0.204	0.394*	0.367*	0.112	0.217
Quad$_{ECC}$	0.500*	0.429*	0.252*	0.298*	0.041	0.444*	0.271*	0.310*	0.054
Ham$_{ECC}$	0.710*	0.405*	0.468*	0.123	0.078	0.349*	0.445*	0.065	0.121
					Females (n=35)				
		Total	**Hip**	**Knee**	**Ankle**	**Total**	**Hip**	**Knee**	**Ankle**
Quad$_{ECC}$		0.426*	0.008	0.389*	0.088	0.336*	0.033	0.351*	0.010
Ham$_{ECC}$		0.310	0.478*	0.182	-0.174	0.275	0.435*	0.163	-0.163
					Males (n=35)				
		Total	**Hip**	**Knee**	**Ankle**	**Total**	**Hip**	**Knee**	**Ankle**
Quad$_{ECC}$		0.193	0.132	0.203	-0.208	0.296	0.150	0.263	-0.124
Ham$_{ECC}$		0.045	0.205	-0.080	0.000	-0.053	0.157	-0.178	0.084

* Significant correlation ($p<0.05$)
† Height$_{STD}$ condition= 0.45±0.0 meters
# Height$_{EQU}$ condition= 0.57±0.7 meters

Table 14 displays the final regression models with standardized coefficients, adjusted R^2 value, and full equation when predicting energy absorption (EA$_{TOT}$, EA$_{HIP}$, EA$_{KNEE}$, and EA$_{ANK}$) with eccentric strength (Quad$_{ECC}$, Ham$_{ECC}$) and sex by strength interactions (sex*Quad$_{ECC}$, sex*Ham$_{ECC}$). During the Height$_{STD}$ condition where all participants landed from a standardized 0.45m height, greater Quad$_{ECC}$ (B= 0.539, R^2 change= 17.2%, $p<0.011$) and Ham$_{ECC}$ (B= 0.380, R^2 change= 4.8%, $p=0.048$) predicted greater EA$_{TOT}$, with the full model explaining 22.7% of the total variance ($p<0.001$). This relationship was not sex-specific. Ham$_{ECC}$ was the sole significant predictor of EA$_{HIP}$ (B= 0.468, $p<0.001$), explaining 20.8% of the variance in EA$_{HIP}$. Quad$_{ECC}$ was the sole significant predictor of EA_{KNEE} (B= 0.535, R^2 change= 7.6%, p=0.012), with a full model R^2 of 7.9% (p=0.037). In each case, the coefficients indicate that greater eccentric torque production ($Quad_{ECC}$ and Ham_{ECC}) predicted greater energy absorption. There were no significant predictors of EA_{ANK}.

Table 14. Final regression models with standardized coefficients, adjusted R^2 value, and full equation when predicting energy absorption (EA_{TOT}, EA_{HIP}, EA_{KNEE}, and EA_{ANK}) with sex, eccentric strength ($Quad_{ECC}$, Ham_{ECC}) and sex by strength interactions ($sex*Quad_{ECC}$, $sex*Ham_{ECC}$).

Dependent Variable	**R^2 value**	**Final Regression Equation**
$HEIGHT_{STD}$ CONDITION		
EA_{TOT}	0.227*	EA_{TOT}= 0.033 + 0.539($Quad_{ECC}$)† + 0.380(Ham_{ECC})† -0.258(sex*$Quad_{ECC}$) - 0.185(sex*Ham_{ECC})
EA_{HIP}	0.208*	EA_{HIP}= -0.009 +0.468(Ham)†
EA_{KNEE}	0.079*	EA_{KNEE}= 0.032 + 0.535($Quad_{ECC}$)† - 0.208(sex*$Quad_{ECC}$) - 0.136(sex*Ham_{ECC})
EA_{ANK}	0.031	EA_{ANK}= 0.031 + 0.252(sex) - 0.142 (sex*$Quad_{ECC}$)
$HEIGHT_{EQU}$ CONDITION		
EA_{TOT}	0.222*	EA_{TOT}= 0.074 + 0.208(sex) + 0.345($Quad_{ECC}$)† + 0.263(Ham_{ECC}) - 0.293(sex*Ham_{ECC})
EA_{HIP}	0.186*	EA_{HIP}= -0.006 + 0.445(Ham_{ECC})†
EA_{KNEE}	0.117*	EA_{KNEE}= 0.049 + 0.424($Quad_{ECC}$)† - 0.245(sex*Ham_{ECC})
EA_{ANK}	0.033	EA_{ANK}= 0.032 + 0.217(sex)

* Significant model R^2 ($p<0.05$)

† Significant regression coefficient ($p<0.05$)

In order to explore the hypothesis that the relationship between eccentric strength and EA would get stronger with greater task difficulty, the same regressions were performed for the $Height_{EQU}$ condition where participants landed from increased heights (average= 0.57±0.07m) based on intra-pair differences in LELM relative to body mass. The prediction models (Table 14) were generally similar to those seen in the $Height_{STD}$ condition. $Quad_{ECC}$ was the sole predictor of EA_{TOT} (B= 0.345, p=0.009) and EA_{KNEE} (B= 0.424, p=0.001) with all included variables explaining an overall R^2=22.2% (p<0.001) and 11.7% (p=0.006), respectively. Ham_{ECC} was the sole predictor of EA_{HIP} (B= 0.445, p<0.001) with the total model explaining 18.6% of the variance (p<0.001). Again, there were no significant predictors of EA_{ANK} during the $Height_{EQU}$ condition. For regression coefficients, standard errors, and t-values of each predictor in the full model when predicting each EA_{TOT}, EA_{HIP}, EA_{KNEE}, and EA_{ANK} with $Quad_{ECC}$, Ham_{ECC}, sex*$Quad_{ECC}$, and sex*Ham_{ECC}, see Appendix H.

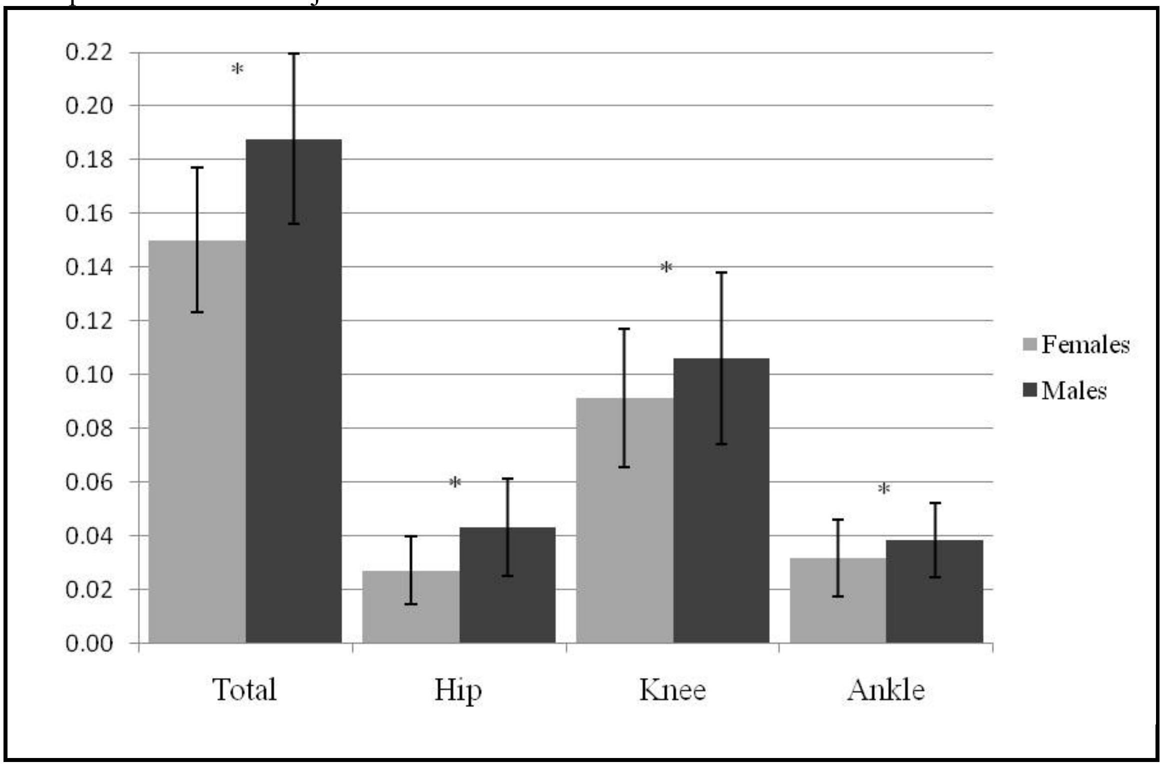
Contrary to the proposed hypotheses, no sex*$Quad_{ECC}$ or sex*Ham_{ECC} interactions were significant for any of the EA models (EA_{TOT}, EA_{HIP}, EA_{KNEE}, or EA_{ANK}) during either landing condition ($Height_{STD}$ or $Height_{EQU}$). This indicates that the strength of the relationship between LELM and eccentric strength ($Quad_{ECC}$ and Ham_{ECC}) was the same for males and females, regardless of the task difficulty examined in this study.

Hypothesis 3: Effect of Equalizing Task Difficulty on Energy Absorption Strategy

To determine the influence of task difficulty on energy absorption strategies, males and females were compared during two landing conditions: $Height_{STD}$ and $Height_{EQU}$. The drop height during the $Height_{EQU}$ condition was determined by calculating the relative difference in LELM between BMI-matched pairs of male-female dyads. The $Height_{EQU}$ calculations (see Appendix B) resulted in an average drop height of 0.57±0.07m, with the increase in height ranging from 0.00-0.23m. An increase of 0.0 meters indicates no difference in the $LELM*PE^{-1}$ ratio (see Appendix B) between the matched female and male. For descriptives of all variables, including the kinetic and kinematic data to aid in interpretation, refer to Tables 5 and 6.

When examining the effect of equalizing the relative task difficulty on energy absorption between males and females ($Height_{EQU}$ in males vs. $Height_{STD}$ in females; Hyp. 3b), the univariate ANOVAs revealed that males absorbed more EA_{TOT} (18.8±3.2 vs. 15.0±2.7; $F_{1,68}$= 28.53; $p<0.001$), EA_{HIP} (4.3±1.8 vs. 2.7±1.3; $F_{1,68}$= 18.05; $p<0.001$), EA_{KNEE} (10.6±3.2 vs. 9.1±2.6; $F_{1,68}$= 4.46; $p=0.038$), and EA_{ANK} (3.8±1.4 vs. 3.1±1.4; $F_{1,68}$= 4.326; $p=0.041$) than females (Figure 5).

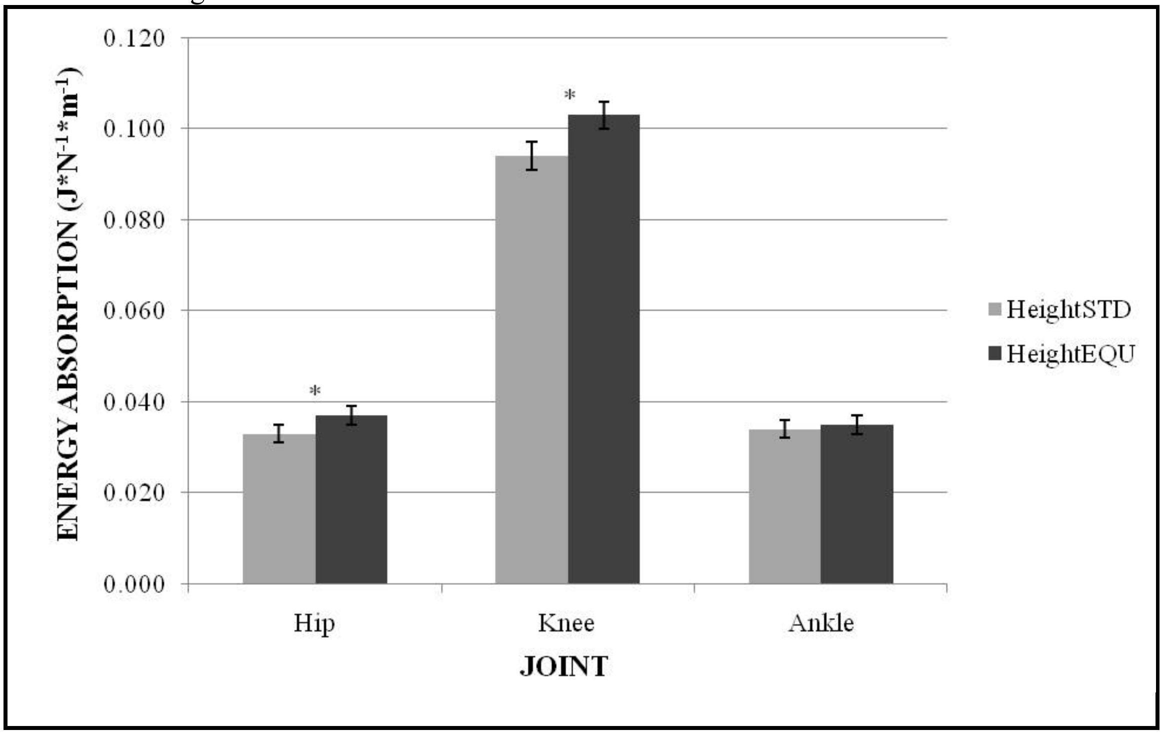
Figure 5. Comparison of males and females when relative task difficulty was equalized ($Height_{EQU}$ in males vs. $Height_{STD}$ in females). Means±SD are displayed for total energy absorption and individual joint contributions.



* Males>Females ($p<0.05$)

The multivariate test comparing males and females on EA_{HIP}, EA_{KNEE}, and EA_{ANK} between $Height_{STD}$ vs. $Height_{EQU}$ conditions identified significant main effects for condition ($F_{3,66}$= 7.21; p< 0.001) and sex ($F_{3,66}$=30.05 ; p<0 .001), but no condition by sex interaction ($F_{3,66}$=0.412, p=0.745). The follow-up univariate tests for condition revealed greater energy absorption about the hip ($F_{1,68}$= 27.54; p<0.001) and knee ($F_{1,68}$= 49.31; p<0.001) when landing from the greater height ($Height_{EQU}$) compared to the lower height ($Height_{STD}$) (Hip: 0.037±0.002 vs. 0.033±0.002 $J*N^{-1}*m^{-1}$; Knee: 0.103±0.003 vs. 0.094±0.003 $J*N^{-1}*m^{-1}$) (Figure 6). However, no differences at the ankle were observed between condition (0.035±0.002 vs. 0.034±0.002 $J*N^{-1}*m^{-1}$; $F_{1,68}$= 2.37, p=0.131).

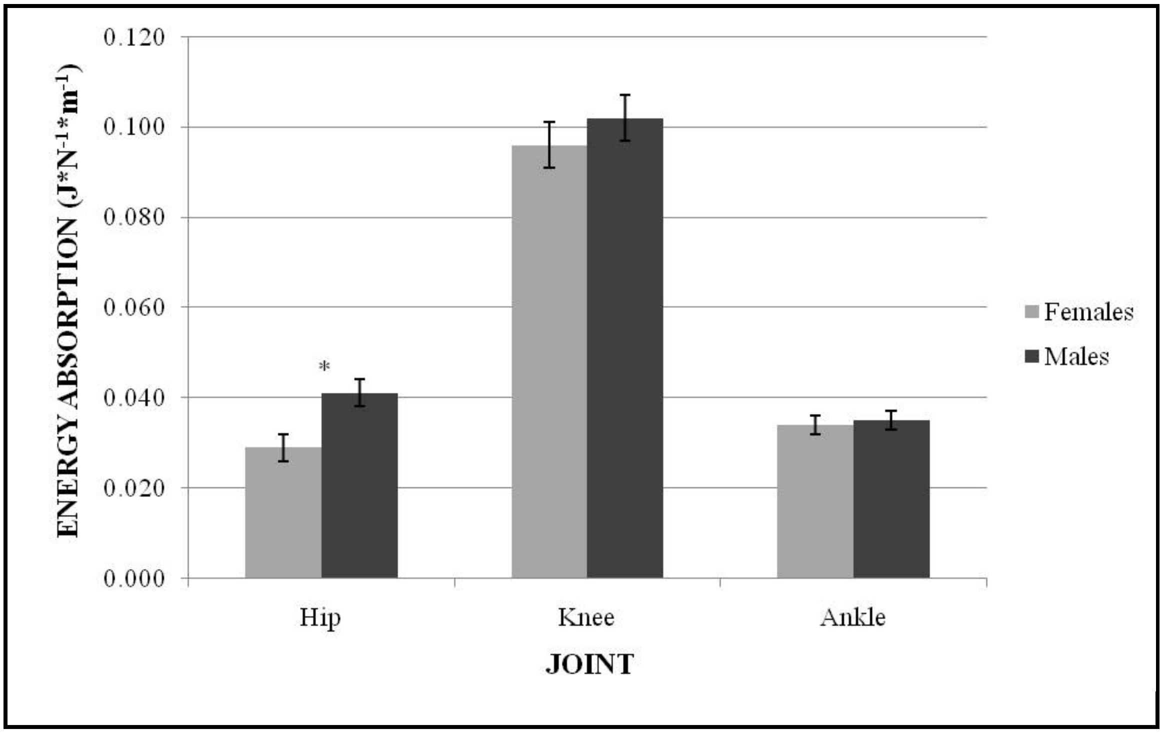
Figure 6. Means±SD for absolute energy absorption at the hip, knee, and ankle when compared between landing conditions.



* $Height_{EQU}$>$Height_{STD}$ ($p<0.05$)

Follow-up univariate tests investigating the main effect of sex revealed that males absorbed more energy about the hip (0.041±0.003 vs. 0.029±0.003; $F_{1,68}$= 10.82; p=0.002) compared to females (Figure 7). No differences between males and females were noted at the knee (0.102±0.005 vs. 0.096±0.005; $F_{1,68}$= 0.956; p=0.332) or ankle (0.035±0.002 vs. 0.034±0.002; $F_{1,68}$= 3.32; p=0.073).

Figure 7. Means±SD for absolute energy absorption at the hip, knee, and ankle when compared between males and females.



* Males>Females(p<0.05)

CHAPTER V

DISCUSSION

The purpose of this study was to investigate the relationships between lower extremity lean mass, strength, and lower extremity energy absorption, and how these relationships may explain sex differences in lower extremity energy absorption strategies. The primary findings were that the relationships between lower extremity lean mass, eccentric quadriceps and hamstring strength, and energy absorption were more pronounced in females, indicating that strength was more important in determining energy absorption strategies for females vs. males. However, the relationships between eccentric strength and energy absorption were not influenced by the relative difficulty of task. In other words, increasing the muscular demands when landing from a greater height did not make strength a more crucial factor in energy absorption. Furthermore, and contrary to the hypothesis, equalizing the task difficulty by accounting for sex differences in body composition did not appear to explain the sex differences in the energy absorption strategies observed. These findings indicate that while females, but not males, demonstrated moderate relationships between lean mass, strength, and energy absorption, strength does not appear to be the determining factor in differentiating energy absorption strategies between males and females during landing. This discussion will focus first on the findings regarding sex differences in body composition and strength and subsequently their relationship with landing energetics. Then sex differences in landing mechanics will be discussed in light of task difficulty.

Sex Differences in Body Composition and Strength

This study confirmed previous findings regarding sex differences in body composition and strength, which is a fundamental tenet to the rationale for this project. Males had, on average, 40% more lean body mass relative to total body mass compared to females, and 42% more lower extremity lean mass, but there were no sex differences observed in the proportion of lean mass distributed in the lower extremities (as expressed by LELM/total body lean mass). The greater amount of total body lean mass in males vs. females in this study is less than the 50-70% difference previously reported in the literature (Malina, 2005; Loomba-Albrecht & Styne, 2009). This may be attributed to the more highly trained population included in the current study (i.e. females having greater lean mass vs. an untrained female population). In regards to the similar relative lean mass distributed in the lower extremities in both males and females, this finding is in agreement with another study investigating sex differences in regional body composition in former Army cadets (Nindl, Scoville, Sheehan, Leone, & Mello, 2002). In that study, females possessed 70% of the absolute lower extremity lean mass of the males, but no differences in lower extremity lean mass when expressed relative to total body lean mass. Along with possessing greater amounts of lean mass, males also demonstrated the expected superiority in eccentric strength of the quadriceps and hamstrings, both in absolute peak torque production and when normalized to body mass. These findings are supported by the literature as well as common observation.

In summary, and as expected, males possessed more lower extremity lean mass and eccentric strength than females, thereby supporting the underlying rationale for this study. As such, the next step was to determine the extent to which these sex differences in strength and body composition explained or contributed to sex differences in energy absorption strategies.

Relationships Between Lean Mass, Strength, and Energy Absorption

Previous literature indicates that the lower amounts of strength in females compared to males may in part explain the stiffer landings performed with less hip and knee flexion (Lephart, et al., 2002), the greater demands on the knee extensors (Salci, et al., 2004; Kulas, et al., 2008; Shultz, Nguyen, Leonard, et al., 2009; Kulas, et al., 2010; Schmitz & Shultz, 2010), and the larger landing forces (Salci, et al., 2004) observed in females compared to males. In support of this theory, greater MVIC strength has been shown to predict greater energy absorption at knee, suggesting that those with greater strength possessed a greater ability to control the body's deceleration about the knee joint (Schmitz & Shultz, 2010).

Since females possess less leg strength and lean body mass, and because production of adequate muscle torques is critical in controlling the body's deceleration during landing and maintaining dynamic joint stability, it was hypothesized that the magnitude of lower extremity lean mass would positively predict the magnitude of energy absorption (Hyp. 1a), but that this relationship would be mediated by the magnitude of eccentric thigh strength (Hyp. 1b). As the primary muscle action in the lower extremity during the deceleration phase of landing is eccentric, it seemed reasonable to expect that eccentric torque measurements would be the best contraction type to examine in relation to energy absorption. As such, eccentric quadriceps and hamstring torques at 180°/s were used to predict lower extremity energy absorption strategies.

When looking at the simple correlations between lower extremity lean mass and eccentric quadriceps or hamstring torques, moderate relationships were present for all subjects. But when stratified by sex, those relationships persisted in females, but not males. This finding is somewhat surprising since previous literature has shown no sex differences in muscle size-strength relationships, albeit during isometric contractions (Maughan, et al., 1983; Akagi, et al., 2009). As lower extremity lean mass was not significantly related to eccentric strength at 180°/s in males, it appears that there may be additional factors that determine peak eccentric torque capabilities at this speed other than lean muscle mass. While it is likely that this sex difference can be attributed to neural mechanisms, sex differences in the relationships between lean mass and dynamic contractions are not well established and need further examination.

Similarly, significant relationships between eccentric quadriceps and hamstring strength and energy absorption about the knee and hip, respectively, were observed in females, but not males. Again, this suggests that the amount of energy that is absorbed during landing is somewhat reliant on eccentric strength capabilities, but only in females.

Despite these significant relationships in females, the results indicate that strength was rarely a mediator of the relationship between LELM and energy absorption. It appears that this was primarily due to the few significant relationships in the individual pathways of the mediation analysis, specifically between 1) lower extremity lean mass and energy absorption and 2) eccentric strength and energy absorption (after controlling for LELM). In females, LELM predicted greater energy absorption only at the knee, yet neither eccentric quadriceps nor hamstring strength mediated the relationship. This essentially means that greater LELM predicted greater EA due to a unique amount of variance that was not shared with eccentric strength. Conversely, if eccentric strength was a mediator, that would have indicated that LELM was predictive of EA only because of its shared variance associated with eccentric strength.

It was noted that eccentric hamstring strength was a mediator of the relationship between LELM and hip energy absorption despite no primary relationship between LELM and hip energy absorption. This suggests that there may have been intervening factors in the relationship between LELM and hip energy absorption (besides strength) which reduced the initial direct effect (Shrout & Bolger, 2002). As such, this situation may be exempt from the first requirement for simple mediation and the nomenclature is slightly modified to denote a significant "indirect effect" (instead of mediating effect) of eccentric hamstring strength on the relationship between LELM and energy absorption (Shrout & Bolger, 2002; Hayes, 2009), although the functional interpretation is still that eccentric strength is an intermediary factor between LELM and hip energy absorption. Plausible intervening factors which would have reduced the initial direct effect would be those most likely related to both LELM and hip energy absorption, such as muscle activation or hip extensor strength (which were not measured).

Although eccentric strength was not found to be a mediator of the relationship between LELM and energy absorption in most cases, previous work has indicated that strength was a positive predictor of energy absorption about the knee in females, but not males (Schmitz & Shultz, 2010). In order to further compare these results to the current results, the extent to which eccentric strength predicted energy absorption (one component of the mediation analysis) was more closely examined. Maximal eccentric hamstring torque predicted 20.6% of the variance in hip energy absorption in females, whereas the aforementioned study (Schmitz & Shultz, 2010) did not find a relationship between maximal isometric hamstring torque and energy absorption about the hip. The disparity between these studies is likely explained by the greater specificity of the hamstring strength measurements in the current study (eccentric vs. isometric) to the action at the hip during drop jump landing. Specifically, the hamstrings are believed to eccentrically contract to assist with control of hip flexion and forward motion of the trunk during rapid deceleration of the body's momentum (Devita & Skelly, 1992), coupled with the loading at the hip necessary to perform a maximal vertical jump (Lees, Vanrenterghem, & De Clercq, 2004).

However, this notion of specificity did not hold for the quadriceps. In the current study, maximal eccentric quadriceps strength explained 12.6% of the variance in knee energy absorption in females, which is not a substantial improvement over the 11% of variance explained by maximal isometric strength (Schmitz & Shultz, 2010). The close similarity in predictive ability of these findings is somewhat surprising considering the preceding discussion regarding specificity of muscle contraction. Because the isometric contractions are likely the least reflective of the muscle action during landing, these findings suggest the possibility that other contraction types or speeds may be related to landing energetics. Specifically, the strength measures which were used in the current study (maximal eccentric contractions at 180°/s) may not have been the most reflective of the type of strength which is most crucial to predicting energy absorption. The rationale for choosing to measure eccentric strength at 180°/s was that it was likely more reflective of the action of the muscles during the deceleration phase of the drop jump compared to strength measures used in previous studies (Lephart, et al., 2002; Bennett, et al., 2008; Schmitz & Shultz, 2010). The action of the hamstrings during landing has been disputed because of the biarticular nature of this muscle group, whereby they shorten distally to flex the knee, but stretch proximally to control trunk and hip flexion (McIntyre et al., 2006; Robertson, Wilson, & St Pierre, 2008). However, as it is less likely that the hamstrings shorten to flex the knee during landing (which is being forced by gravity), but rather that they contract eccentrically to perform controlled trunk and hip flexion, eccentric hamstring strength was chosen for this analysis.

Because there were sex differences in the relationship between eccentric strength and energy absorption, it is plausible that there may also be sex differences in the strength measurements that are most related to energy absorption. Indeed, preliminary inspection of additional strength data collected from these subjects (isometric and concentric contractions) revealed that the eccentric contractions at 180°/s were not the most highly correlated with energy absorption and also that sex differences exist in the types and speeds of contractions that are best related to landing energetics (Appendix E). While this finding is not wholly understood right now, it may be partially explained by sex differences in muscle function attributed to differences in the underlying physiological characteristics of the muscle and could affect performance of dynamic contractions.

Some evidence suggests that females have a greater reliance on "slow" Type I muscle fibers in the quadriceps compared to males (Simoneau & Bouchard, 1989; Staron et al., 2000). Functionally, this underlying physiological characteristic may help explain sex differences in torque-producing capabilities in the experimental setting which have some implications for the current findings. For example, females have been shown to have longer electromechanical delay (Winter & Brookes, 1991) than males and a lessened ability to perform fast speed isokinetic contractions compared to males (L. E. Brown, Whitehurst, Gilbert, & Buchalter, 1995), thus resulting in impaired torque producing abilities compared to the slower speed contractions. If females have greater difficulty producing eccentric torques at faster speeds, it could provide a partial explanation for the stronger relationship between eccentric strength and energy absorption in females. It could be that the ability to produce larger eccentric torques at 180°/s has more important implications for the ability to produce adequate eccentric torques to control deceleration and absorb energy during landing in females, whereas males do not have a problem producing adequate eccentric torques during either condition (strength testing or landing). Hence, eccentric torque was not a significant determinant of energy absorption in males.

The weak relationships between eccentric strength and energy absorption found in this study may also be due to the methodologies employed, specifically the attempt to relate a peak torque measurement captured at a specific knee flexion angle with a landing maneuver synthesized over a range of knee flexion angles (i.e. foot contact to peak knee flexion). This contention has been suggested by previous investigators who were unable to relate peak strength measures to peak anterior knee shear force (Bennett, et al., 2008) despite a strong rationale to support their relation. In order to address this limitation, perhaps a more comprehensive measurement of strength such as average torque (throughout range of motion) or work (the area under the joint power curve) would relate better to energy absorption during landing because of its closer specificity to the energetics measurement.

There also remains the possibility that these contractions did not relate well because of the limitations associated with the inherent difficulty in performing fast eccentric contractions in a dynamometer, regardless of sex. Test-retest reliability of the non-normalized peak torque values for a subset of the subjects in the current study (n=15) ranged from 0.91-0.98 ($ICC_{2,k}$) (Appendix E), which is very similar to a published study using a similar protocol (R. C. Li, Wu, Maffulli, Chan, & Chan, 1996). This suggests that the subjects were able to perform the contractions with reasonable consistency from day-to-day, but does not address a limitation of all strength testing protocols: difficulty in assessing the validity of the measurements. In this sense, validity refers to the ability to capture the true force-producing capability of the available musculature, which requires maximal effort and excellent execution of the muscle contractions. While every effort was made to ensure that the subjects were well familiarized and that they performed consistently and maximally across trials, this was the extent to which the validity of the peak torque measurements could be evaluated, specifically since the muscle size-strength relationship during dynamic contractions is unclear. As the muscle size-strength relationship has been shown to be linear during maximal isometric contractions (Bamman, et al., 2000), and equal between men and women (Maughan, et al., 1983; Akagi, et al., 2009), examining these same relationships during concentric and eccentric contractions at various speeds could serve to provide reference data against which validity may be assessed. It may also elucidate possible inconsistencies in the relationship between maximal force production and available muscle mass. Specifically, these inconsistencies may represent non-linear relationships between lean mass and dynamic strength or could point to inaccuracies in maximal force producing capabilities due to difficulties with performance of the strength testing protocol. Either way, this information is important to parse out and could serve to better inform future attempts to relate lean mass to strength during dynamic contractions.

Qualitatively, it was noted that subjects had greater difficulty with the faster contractions, specifically the eccentric quadriceps contractions at 180°/s, which is evidenced by the lowest reliability of all contraction types and speeds ($ICC_{2,k}$= 0.91, SEM= 16.0 Nm). Inconsistencies in performance of this strength measurement may be partially responsible for the lower than expected relationships with energy absorption. It should also be noted that the maximal eccentric strength testing protocol used in this study caused substantial muscle soreness in the hamstrings. Although a subjective scale (e.g. VAS) was not used to quantify the amount of soreness in each subject, the amount of soreness seemed to vary substantially across subjects (i.e. some people did not experience any soreness while others experienced debilitating pain). While the majority of subjects reported feeling completely recovered when reporting for data collection approximately one week following familiarization, some were not able to produce the same peak torque levels achieved during familiarization. When this occurred, the subject was asked to come back for strength testing a few days later. It was not noted that this occurred more often in males or females. Because the overwhelming majority of subjects performed better on their testing day compared to familiarization day, it was difficult to assess whether the peak torques produced during the testing day were truly reflective of the force-producing capabilities of the available amount of lean mass. When a particular subject did not demonstrate the expected improvement in peak torque production (from familiarization to testing day), it was difficult to assess the reasons for the lack of improvement. The two most likely reasons were that there was some lingering muscle dysfunction (but no muscle soreness) or that the more highly-trained subjects actually performed closer to their true peak torque production on the familiarization day and therefore were not subject to the large learning curve leading to vast improvement on the test day. However, it was not possible to differentiate between these two situations. Hence, the quality of data could only be judged by the perception of maximal effort and the subject's consistency across trials (as assessed by the coefficient of variation). It is also acknowledged that 180°/s is substantially slower than the average angular velocity of knee flexion during landing (~600-700°/s) (Decker, et al., 2003); however, since 180°/s is fastest speed in the literature which shows good reliability (ICC>0.80) (Perrin, 1986; Tredinnick & Duncan, 1988; R. C. Li, et al., 1996) during similar protocols, this testing speed was chosen as the best balance between reliability and external validity in regard to muscle function during landing.

Another relevant methodological limitation acknowledged in this study is that only knee flexor and extensor strength were measured. Because the purpose of this study was to examine the effect of eccentric strength on lower extremity energy absorption as it relates to knee injury risk, it was fundamentally important to measure the muscles that directly control knee function. However, as energy absorption of the hip and ankle were also of interest, it is likely that measuring the strength of the primary musculature at those joints, specifically the primary hip extensors (gluteals) and plantar flexors (gastrocnemius-soleus complex), would have improved the relationships between strength and joint-specific energy absorption. Further, since the muscles do not act in isolation (as tested in this study), in the future, capturing an "index" of lower extremity strength would likely provide a more comprehensive assessment of the global force-producing capabilities and may further improve the ability to relate strength and energy absorption. Additional work should focus on developing valid field-based multi-joint strength measurements which more closely mimic the function of the muscles during dynamic weight-bearing activities. Such field-based tests could potentially reduce the need for burdensome equipment and allow for large-scale screening of strength capabilities with the goal of furthering our understanding of the role of functional strength in lower extremity neuromechanical strategies.

In summary, the current findings concur with previous literature which has shown significant relationships between strength and energy absorption only in females. Additional information gained from this study was the significant relationship between lower extremity lean mass and eccentric strength in females, but not males. While eccentric strength did not appear to be a strong mediator of the relationship between lean mass and energy absorption, individual relationships among these variables merits further examination using other measurements of thigh strength. Although the source of these sex differences is unclear at this time, it appears that the underlying mechanisms which determine energy absorption capabilities are different between males and females. As such, it appears prudent to perform separate analyses for males and females when examining relationships between strength and energy absorption so as to not obscure any potential relationships. Further, it is clear that more work is needed to elucidate the types of contractions best related to energy absorption and also to identify the mechanisms that may be responsible for potential sex differences in said relationships.

Influence of Task Difficulty on Relationships between Strength and Energy Absorption

Previous literature indicates that quadriceps strength (MVIC) predicts energy absorption in females but not males (Schmitz & Shultz, 2010). These authors suggested that stronger relationships in females were likely due to greater relative task difficulty for the females as they possess less strength for a given body mass, thereby requiring greater muscular effort to decelerate the body's momentum from a given height, compared to a similarly sized male. In the current study, a multiple linear regression analysis was used to test this theory and determine whether the relationship between eccentric strength and energy absorption was greater in females due to the greater relative task difficulty for females compared to males. This hypothesis was further tested by equalizing the relative task difficulty (i.e. increasing the drop height) between males and females based on sex differences in lower extremity lean mass, and then examining relationships between strength and landing energetics again at this higher (i.e. more demanding) height.

The hypothesis that the relationships between strength and EA would be stronger in females due to the greater relative difficulty of the landing task at the standard (non-normalized) height was not supported. That is, neither sex nor interactions of sex with eccentric quadriceps or hamstring strength were significant predictors in the model. This was an unexpected finding since previous literature (Schmitz & Shultz, 2010) suggests that strength is a stronger predictor of energy absorption in females versus males due to greater relative task difficulty for females. Additionally, the expectation that the relationships between strength and EA would get stronger because strength would become a more important factor in absorbing energy as the task became more difficult, was not supported for females or males.

Contrary to the hypotheses, differences in task difficulty between sex or condition did not appear to influence the relationships between eccentric strength and energy absorption. When analyzing energy absorption during the more difficult Height_{EQU} condition (which was on average 38% higher than the Height_{STD} condition), the correlations and regression models were very similar to what was observed when analyzing the $Height_{STD}$ condition. These results would indicate that sex differences in relative task difficulty were not responsible for the previously-observed sex differences in relationships between strength and energy absorption.

Perhaps one explanation for this finding is that neither landing task was challenging enough to make eccentric strength a determining factor in energy absorption capabilities for males or females. As the previous study (Schmitz & Shultz, 2010) examined these relationships in a recreationally-active population, it is possible that the highly athletic population in the current study possessed greater strength and landing skill compared to the recreationally-active population, and was therefore able to perform the landing task with greater relative ease, thus reducing the importance of maximal strength. In previous studies, recreationally-active subjects successfully landed from various heights up to 1.03-1.28 meters (McNitt-Gray, 1993; Zhang, et al., 2000; Zhang, et al., 2008) which is substantially higher than the average drop height of 0.57 m during the current study. This suggests that the current subjects were not challenged near their full capabilities.

If the larger drop height did not require a large amount of the available torque-producing capabilities of the participants, this may have resulted in a further mismatch of specificity of muscle contraction type and energy absorption, especially since the analyses in this study were based on maximal eccentric torque production. As it is unlikely that one's maximum force capabilities are utilized when performing a landing maneuver, this is another plausible explanation for why maximal eccentric strength was not a strong predictor of energy absorption. Instead, how one utilizes this strength via muscle activation strategies may be an intermediary factor between maximal eccentric strength and energy absorption. It may be that the muscle activation level and timing may aid in the ability to explain energy absorption capabilities.

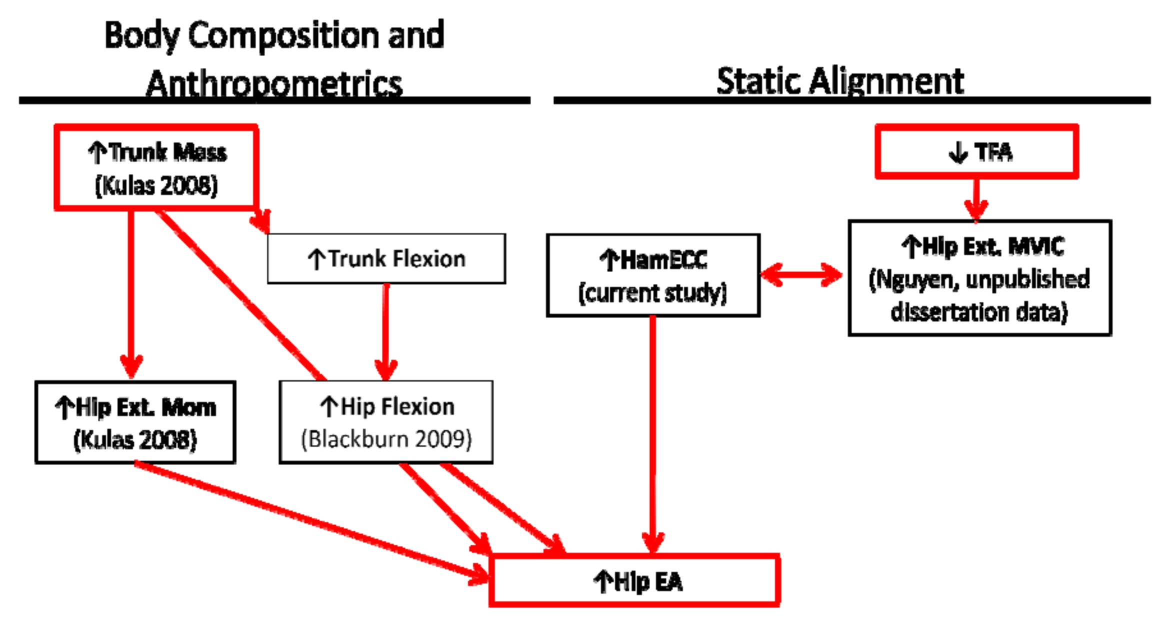
Females have been shown to perform landing maneuvers with greater quadriceps muscle activation amplitudes (Malinzak, et al., 2001; Chappell, et al., 2007; Shultz, Nguyen, Leonard, et al., 2009), which can in part be explained by a lower amount of maximal isometric strength (Shultz, Nguyen, Leonard, et al., 2009). As lower MVIC strength has also been related to less energy absorption capabilities about the knee (Schmitz & Shultz, 2010), collectively this may suggest a possible relation between muscle activation amplitudes and energy absorption. Because surface electromyography (sEMG) can be a useful tool for making inferences about muscle activity in relation to maximal isometric torque production (Woods & Bigland-Ritchie, 1983), a more comprehensive analysis of the influence of maximal eccentric strength producing capabilities (via strength testing) plus an inference of the relative use of those capabilities (via sEMG) would most likely improve our understanding of the influence of strength on energy absorption strategies.

Apart from improving the most appropriate measure of strength and muscle function as they relate to energy absorption, it is likely that other factors also influence energy absorption strategies, as much of the variance in energy absorption remained unexplained by strength. Although the models were statistically significant, it is difficult to argue that the ability to explain 20.6% and 12.6% of the variance in hip and knee energy absorption, respectively, provides a meaningful representation of energy absorption capabilities. While it is difficult to tell what additional factors might be from the current data, only one other study has attempted to predict energy absorption capabilities. Shultz et al (Shultz, et al., 2010) found that a combination of greater anterior knee laxity and general joint laxity predicted greater energy absorption about the knee, explaining approximately 21% of the variance. However, even collectively, strength and laxity do not account for a large majority of the variance in energy absorption (~37%), which is still only a moderate effect.

In the attempt to reveal additional factors which may account for the unexplained variance in energy absorption during this study, further review of the literature revealed transitive inter-relationships between body composition, static alignment, knee joint laxity, strength, and energy absorption in females. Liberal use of this literature suggests that the addition of select anthropometric, alignment, and laxity variables would likely improve the energy absorption predictions and collectively explain a larger proportion of the variance in energy absorption. Figures 8 and 9 display a theoretical framework proposing anthropometric characteristics which may predict energy absorption capabilities of the hip and knee, respectively.

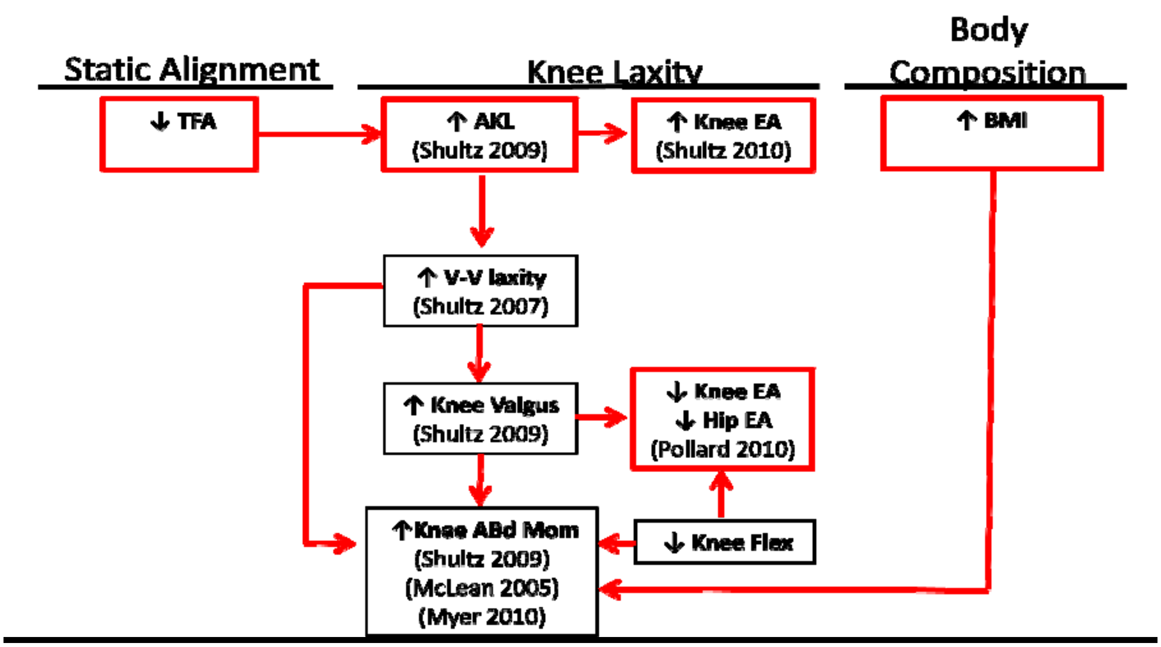
It appears that trunk mass may have a significant influence on energy absorption strategies at the hip. Specifically, a greater relative mass of the HAT segment results in a larger moment of inertia and may explain the greater hip flexion and extensor moments at the hip during landing as an individual attempts to control the forward and downward motion of the segment (Devita & Skelly, 1992). This is supported by previous work that shows that artificially loading the trunk resulted in greater trunk flexion, greater hip angular impulse, and a strong trend towards greater hip energy absorption (Kulas, et al., 2008). Additionally, greater trunk flexion is accompanied by greater hip flexion (Blackburn & Padua, 2008), which also likely result in greater hip energy absorption. Collectively, these findings support an influence of trunk mass on hip energy absorption capability.

Figure 8. Theoretical framework proposing anthropometric characteristics which may predict energy absorption capabilities of the hip.



In addition, several lower extremity alignment characteristics together may influence hip extensor torque, which would likely influence energy absorption capabilities at the hip. Specifically, a smaller tibiofemoral angle is related to greater hip extensor MVIC (i.e. gluteal) (r=-0.307) (Nguyen, unpublished data). This measurement likely relates well to the Ham_{ECC} strength measurement in the current study due to the extensor torques produced at the hip with eccentric action of the hamstrings. As Ham_{ECC} was a positive significant predictor of energy absorption at the hip, these relationships may in turn suggest tibiofemoral angle as a relevant anatomical alignment characteristic which influences energy absorption at the hip.

Figure 9. Theoretical framework proposing anthropometric characteristics which may predict energy absorption capabilities of the knee.



Interestingly, a smaller tibiofemoral angle is also related to greater anterior knee laxity (Shultz, Nguyen, & Levine, 2009). As mentioned previously, greater anterior knee laxity has been shown to predict greater knee energy absorption in females, explaining 11% of the variance (Shultz, et al., 2010). Since greater anterior knee laxity is also related to greater varus-valgus laxity (r=0.695) (Shultz, et al., 2007), it is plausible that frontal plane laxity also influences energy absorption capabilities at the knee. While the relationship between frontal plane laxity and energy absorption has not been examined directly, greater varus-valgus laxity has been associated with a greater dynamic knee valgus posture during drop jumping (Shultz & Schmitz, 2009). In turn, greater knee valgus during landing has been implicated as a consequence of landing with less hip and knee energy absorption (Pollard, Sigward, & Powers, 2010). Finally, and particularly pertinent to the current study, recent evidence implicates body composition in high school age females as a factor which predisposes one to greater knee abduction loads (Myer, Ford, Khoury, Succop, & Hewett, 2010), which is also related to increased varus-valgus laxity (Shultz & Schmitz, 2009) and valgus postures during landing (McLean, Huang, & van den Bogert, 2005; Shultz & Schmitz, 2009).

Although these relationships are transitive, this theory does suggest the possibility that alignment, body composition, and frontal plane laxity may explain additional variance in knee energy absorption. Additionally, since the direct relationship between anterior knee laxity and knee energy absorption is positive while the aforementioned transitive relationships between body composition, varus-valgus laxity, knee valgus posture and loading, and energy absorption during landing are negative, this may additionally imply that anterior knee laxity and varus-valgus laxity are explaining separate phenomena which predict energy absorption capabilities. While this theoretical framework needs additional development and will be modified with further research, it does indicate that these theoretical relationships deserve further investigation in the attempt to explain a functionally relevant amount of lower extremity energy absorption capabilities.

Based on the current study, strength does play a role, but since the relationships with other relevant factors has barely been examined, the magnitude of its importance relative to the other pertinent factors cannot be determined at present time. What is clear from this theoretical framework is the multifactorial nature of the determinants to lower extremity biomechanics. As such, it does not appear that continuing to investigate singular factors will aid in our understanding of the determinants of energy absorption.

In summary, weak to moderate relationships between eccentric strength and energy absorption about the hip and knee were observed, independent of sex. However, the strength of these relationships did not appear to be influenced by task difficulty, which was contrary to the hypothesis. As such, it appears that sex differences in strength, thus relative task difficulty, are not responsible for sex differences in energy absorption capabilities during landing. With the large amount of variance unaccounted for by sex and strength, additional work is needed to elucidate the factors which may collectively explain energy absorption capabilities.

Sex Differences in Energy Absorption Strategies

In order to confirm the previous findings regarding sex differences in energy absorption strategies upon which the rationale for the current study is based, the following section describes the energy absorption strategies utilized by the males and females during a standardized task. These findings will be compared to previous literature with the goal of aiding in the interpretation of the findings regarding the effect of equalizing task difficulty in the discussion to follow. Descriptives are presented in Table 5.

In the current study, males absorbed more total energy across the lower extremity joints than females when landing from a standardized height. This is a novel finding as the 3 studies which have compared landing energetics in males and females reported no sex differences in total energy absorption during double-leg landings (Decker, et al., 2003) and drop jumps (Schmitz & Shultz, 2010; Shultz, et al., 2010). When examining individual joint contributions, the current study found that males absorbed more energy at the hip, but there were no sex differences at the knee or ankle, suggesting that the greater energy absorbed at the hip is driving the sex differences in total energy absorption observed. These findings are in partial agreement with previous reports that males absorb more absolute energy about the hip during a terminal landing (Decker, et al., 2003) but are contrary to findings that females absorb more absolute energy about the knee during terminal (Decker, et al., 2003) and non-terminal (Schmitz & Shultz, 2010; Shultz,et al., 2010) landings. Further, the lack of sex differences at the knee is contrary to the proposed hypothesis (Hyp 3a).

To aid in interpretation of the energetics strategies utilized by males and females in this study, relative joint contributions to total energy absorption (%) are provided in Table 15.

Table 15. Mean±SD for relative joint contributions to total energy absorption during both landing conditions separately and also when the task was equalized between BMI-matched males and females ($Height_{EQU}$ in males vs. $Height_{STD}$ in females). Effect sizes are provided for sex and condition.

RELATIVE JOINT CONTRIBUTIONS (% TOTAL ENERGY ABSORPTION)				
		Females (n=35)	**Males (n=35)**	**Between-Sex Effect Size**
$Height_{STD}$	Hip	18.1±8.4	22.4±9.6	0.45
(0.45±0.00 meters)	Knee	60.5±10.8*	56.0±10.6	0.43
	Ankle	21.3±9.3	21.6±7.0	0.04
$Height_{EQU}$	Hip	18.7±7.8	23.2±9.5†	0.48
(0.57±0.07 meters)	Knee	61.3±10.1*	56.0±10.6	0.50
	Ankle	20.0±8.9	20.8±7.0	0.10
Between-Condition Effect Size	Hip	0.07	0.08	
	Knee	0.07	0.00	
	Ankle	0.14	0.12	
Equalized Condition	Hip	18.1±8.4	23.2±9.5†	0.53
	Knee	60.5±10.8*	56.0±10.6	0.43
	Ankle	21.3±9.3	20.8±7.0	0.08

* Females > Males ($p<0.05$)
† Males > Females ($p<0.05$)

When examining sex differences in the relative joint contributions to total energy absorption during the standardized condition, females absorbed a greater proportion of energy about the knee compared to males, while males absorbed a greater proportion of energy about the hip than females. The findings at the knee are consistent with the only other studies that used a drop jump task (Schmitz & Shultz, 2010; Shultz, et al., 2010). While this has been suggested to be reflective of hip extensor weakness (Zhang, et al., 2000; Pollard, et al., 2010), we only examined thigh strength in the current study so this conjecture cannot be examined at the present time.

In the current study, the knee contributed the most to energy absorption in both females and males. This was followed by the hip then ankle in males, while the ankle was the #2 shock absorber in females. The order of relative joint contributions found in this study is similar to what is reported in previous drop landing literature and indicate that the subjects used soft landing strategies (Decker, et al., 2003; Zhang, et al., 2008; Norcross, Blackburn, Goerger, & Padua, 2010). The magnitude of relative energy absorbed about the knee is on average 15-20% larger than was reported in the drop landing studies (Decker, et al., 2003; Zhang, et al., 2008) and upwards of 40% greater than the drop jump studies (Schmitz & Shultz, 2010; Shultz, et al., 2010). However, the current findings are very similar to a study which used a 30cm forward drop jump landing task and reported relative joint contributions from the hip, knee, and ankle at 19.3%, 52.8%, and 28.2%, respectively (Norcross, et al., 2010). It is plausible that the small discrepancy at the knee and hip can be attributed to the more vertical nature of the drop jump task in the current study compared to the greater horizontal component of the comparative task.

Another difference in the current study is the highly skilled and trained jumpers used in the current study compared to the recreationally-active subjects typically used in previous studies (Zhang, et al., 2000; Decker, et al., 2003; Zhang, et al., 2008; Norcross, et al., 2010; Schmitz & Shultz, 2010; Shultz, Schmitz, Nguyen, & Levine, 2010). The overwhelming majority of subjects in the current study were experienced athletes who are currently or were formerly engaged in formal training consisting of instruction in safe landing technique (i.e. those that minimize joint loads) and optimal drop jumping performance. In addition to training, these athletes regularly performed similar activities in their sports. Hence, it is possible that the current findings are reflective of the strategy utilized to optimize the absorption, storage, and subsequent use of energy to create explosive movements for maximal performance. Additionally, because the subjects were not instructed on how to perform the landing, but rather in the desired outcome of performing the largest vertical jump possible, this may have added additional inter-individual strategies among the subjects. Previous work suggests that maximal jump height can be achieved by either a large amount of negative work and smaller knee flexion or by doing less negative work and using more knee flexion during the countermovement (Moran & Wallace, 2007). The choice of strategy employed to attain maximal jump height has not been shown to be sex specific, and it is likely that a mixture of strategies was employed by the subjects in the current study. Thus, no overwhelming group strategy was observed, which may have tempered the changes in energy absorption between conditions and may also have diluted or eliminated sex differences in energy absorption strategy.

Effect of Equalizing Relative Task Difficulty on Energy Absorption Strategies

To further test the theory that sex differences in landing biomechanics are partly attributed to females possessing less lean mass and strength than males (Shultz, Nguyen, Leonard, et al., 2009; Schmitz & Shultz, 2010), the relative task difficulty was equalized between BMI-matched males and females by increasing the drop height for males. As such, it was hypothesized that equalizing the relative task difficulty between males and females based on the amount of lower extremity lean mass would result in similar energy absorption strategies (Hyp 3b), specifically at the knee. Additionally, each female was also asked to land from her matched male's equalized height to test the hypothesis that the change in energy absorption strategy from the lower to the higher drop height would be larger in females compared to males due to the greater overall difficulty for females (Hyp 3c). These hypotheses were not upheld.

Equalizing the drop height for males resulted in greater overall energy absorption, but also a proportional increase across all joints. The increase in energy absorption with greater drop height was consistent with previous studies that have investigated the effects of increased landing heights on energy absorption (McNitt-Gray, 1993; Zhang, et al., 2000; Zhang, et al., 2008). Because females absorbed less energy during the initial standardized condition, comparing the males' equalized condition resulted in even greater sex differences in energy absorption.

In order to further examine the effect of equalizing and increasing the task difficulty between males and females, between-sex effect sizes for the relative joint contributions to total energy absorption were also examined (Table 15). The findings from this post hoc analysis were similar to the absolute joint energetics in that the between-sex effect sizes when males and females were compared during the equalized condition ($Height_{EQU}$ in males vs. $Height_{STD}$ in females), which were virtually identical to those when males and females were compared during non-equalized conditions (Males vs. Females during $Height_{STD}$ and $Height_{EQU}$). This suggests that although males absorbed more total energy when landing from a greater height, the energy absorption strategy at the knee did not change at all, which was contrary to the hypothesis. In fact, relatively speaking, the relative joint contributions shifted a negligible amount (0.0-0.8%

at the hip and ankle). Likewise, the females experienced negligible changes (0.8-1.3%) in energy absorption strategy when comparing the two heights.

The response in females was particularly unexpected since the $Height_{EQU}$ condition represented an especially exaggerated increase in task difficulty compared to the males; hence, larger changes were expected in the females from $Height_{STD}$ to $Height_{EQU}$. Rather, males and females increased their energy absorption in a nearly identical fashion in regard to individual joint contributions. This is evidenced by the similar between-condition effect sizes for males and females (Table 5 and Table 15).

Another unexpected finding was the absence of a "shift" in relative joint contributions to total energy absorption from the distal musculature (i.e. ankle) at the lower height to the larger, more proximal musculature (i.e. hip) at the higher height, which has been previously documented (Zhang, et al., 2000; Zhang, et al., 2008). As the primary sex difference in energy absorption at the standardized height was expected to be about the knee, it appears that the multivariate design used was inadequate as it did not allow for examination of individual joint contributions. Although this was not a part of the original analysis, examination of the energy absorption performed at each joint (without the limitations imposed by the multivariate design) provides an opportunity to more closely examine and classify the collective energy absorption strategy. Again, this was examined via post hoc calculations of between-condition and between-sex effect sizes for each joint (Table 5). It has been previously suggested that larger muscle groups may possess a greater capacity for energy absorption (Zhang, et al., 2000), which is supported by findings that greater muscle cross-sectional area is related to greater passive energy absorption capabilities at the calf (Ryan et al., 2009) so it was reasonable to expect that greater energetic demands would be better met with larger muscle groups. However, this conjecture was not supported in the current study. Although the females demonstrated a trend towards this shift by decreasing energy absorption about the ankle during $Height_{EQU}$, neither the hip nor the knee demonstrated significant increases. Since the drop jump task is typically characterized by larger contributions from the hip compared to the drop landing (Shultz , Tritsch, Montgomery, & Schmitz, In Review), this distal-to-proximal effect may have been negated during the current task since the subjects were already utilizing their hip to absorb energy at the lower height.

Collectively, these findings were unexpected, based on the hypothesis that greater energy absorption demands at the knee in females were likely reflective of a greater difficulty in decelerating the body safely during landing. Specifically, an increase in energy absorption about the knee was expected during the more difficult task. Further, we would have expected an even greater sex difference in energy absorption when males and females both landed from the greater height, which was not met.

As energy absorption represents a global strategy which takes into account joint motion, loads, and velocities, the underlying mechanisms by which energy absorption may be modulated are numerous. That is, while the global strategy may not have changed, examining other biomechanical factors may help explain and provide some significance to the findings. For example, joint range of motion may explain larger energy absorption values (Zhang, et al., 2000; Decker, et al., 2003). However, when examining the underlying biomechanics (i.e. kinematics, kinetics, joint torsional stiffness), there were no other biomechanical factors which appeared to “drive” the energetic strategies. Consistent with the energetics findings, the between-condition effect sizes for other biomechanical factors were similar for males and females. Hence, it appears that the accommodation strategy was the same for males and females.

The most likely explanation for the lack of sex differences in knee energy absorption at either height is that the females in this study were more highly trained, experienced, and stronger than the recreationally-active college students used in previous studies (Zhang, et al., 2000; Decker, et al., 2003; Zhang, et al., 2008; Norcross, et al., 2010; Schmitz & Shultz, 2010; Shultz, et al., 2010). It seems likely that including this population narrowed the expected sex differences in body composition and strength which would have decreased the magnitude of task equalization needed between males and females. As strength is likely not the sole factor in determining energy absorption, other factors which are difficult to quantify, such as coordination, likely played a role as well. Specifically, the high athleticism and training history (i.e. injury prevention strategies and performance optimization) of the females in this study likely served to minimize the previously-observed sex differences in energy absorption strategy, ultimately providing a partial explanation for the lack of expected sex differences at the knee during the standardized condition.

Collectively, these factors could explain the absence of some of the expected sex differences during the $\text{Height}_{\text{STD}}$ condition which are typically observed and attributed to strength differences (e.g. greater vertical ground reaction forces, peak knee extensor moments, knee energy absorption, etc.). Further, the increased drop height during the

$Height_{EQU}$ condition may have still been well within their strength capabilities; thus, their accommodation strategy looked similar to the males. This may point to an inadequacy in the matching of relative demands used in this study. Given that eccentric strength correlated slightly better with energy absorption than lower extremity lean mass, matching relative difficulty on the basis of strength rather than lower extremity lean mass may better capture the true functional differences in relative task difficulty experienced by males and females.

Another observation in this study that became evident during the matching scheme was that matched males and females who had larger BMIs typically had larger drop height differences ($Height_{EQU}$-$Height_{STD}$). Because the equalized condition was essentially based on the ratio of lower extremity lean mass to body mass, and larger differences in lower extremity lean mass result in larger equalized drop heights, this confirmed previous findings that larger BMIs are primarily attributed to greater lean mass in males and greater fat mass in females (Loomba-Albrecht & Styne, 2009). The subjects in this study represented a seemingly large range of BMIs: 19.0 to 29.2 kg/m^2, which accordingly represented a large range of lower extremity lean mass values: 11.3-19.8 kg in females and 16.9-27.3 kg in males. As the difference in lower extremity lean mass relative to body mass between males and females with larger BMI's was often disproportionately larger than the differences between males and females with smaller BMIs, this also resulted in disproportionately larger drop heights during the $Height_{EQU}$ condition for those with larger BMIs. There is a possibility that those males and females who had to drop from higher heights (due to a higher BMI) experienced larger changes in energy absorption in response to their greater drop heights. However, because of the range of drop heights used across all subjects (0.45-0.68 meters), it is possible that combining those with lower drop heights with the larger ones obscured the findings. In other words, smaller increases in drop height were likely met with smaller changes in energy absorption strategy which may have diluted the larger changes in energetics experienced by those who had larger increases in their drop height. To explore this possibility, between-condition effect sizes were computed for males and females when split into groups based on above- and below-average (mean $Height_{EQU}$= 0.57±0.07m) drop heights during the $Height_{EQU}$ condition (Table 16). Additionally, the delta change between landing conditions ($Height_{EQU}$-$Height_{STD}$) was calculated for each group, and the subsequent between-sex effect sizes were examined.

Table 16. Absolute energy absorption for males and females when split into above- and below-average $Height_{EQU}$ drop height (mean = 0.57±0.07m) groups. Mean±SD are provided for each landing condition as well as the delta change from ($Height_{EQU}$-$Height_{STD}$). Between-condition and between-sex effect sizes are also provided.

ENERGY ABSORPTION (J x N^{-1} x m^{-1})*

		Above Average (ht= 0.62±0.03m)		**Below Average (ht=0.51±0.04m)**		**Between Sex Effect Size**	
		Females (n=18)	**Males (n=17)**	**Females (n=17)**	**Males (n=18)**	**Above Avg**	**Below Avg**
$Height_{STD}$	Hip	2.7±1.0	3.6±1.5	2.7±1.5	4.1±1.9	0.58	0.75
	Knee	8.6±1.8	9.6±3.1	9.7±3.2	9.9±2.2	0.31	0.10
	Ankle	3.4±1.3	4.2±1.3	2.9±1.6	3.3±1.0	0.60	0.34
$Height_{EQU}$	Hip	3.2±1.2	4.2±1.4	2.9±1.6	4.4±2.1	0.71	0.71
	Knee	9.7±2.2	10.7±3.9	10.3±2.9	10.5±2.6	0.27	0.08
	Ankle	3.5±1.4	4.2±1.6	2.9±1.5	3.5±1.2	0.43	0.55
Delta Values	Hip	0.5±0.4	0.6±0.5	0.2±0.5	0.3±0.9	0.31	0.07
	Knee	1.0±0.9	1.1±1.3	0.6±0.6	0.6±1.0	0.08	0.04
	Ankle	0.2±0.6	0.0±0.8	0.0±0.3	0.3±0.7	0.18	0.46
Between Condition Effect Size	Hip	0.38	0.42	0.12	0.12		
	Knee	0.47	0.28	0.23	0.23		
	Ankle	0.13	0.03	0.03	0.22		

* Values are expressed x10^2

Males and females with above-average drop heights and BMIs (0.62±0.03m, 24.1kg/m^2) had greater increases in energy absorption when going from the standardized to equalized heights than those with below average drop heights and BMIs (0.51±0.04m, 22.6 kg/m^2), which may be expected simply due to the greater drop heights and generation of kinetic energy. Of particular interest is the moderate between-condition effect size noted at the knee in females (d= 0.47) in the above average group, suggesting a trend in energy absorption at the knee in response to the greater increase in drop height. However, when further examining the effect sizes for sex differences between conditions (i.e. delta) at the knee, the effect size was small (d= 0.08). Sex differences were more apparent at the hip (d= 0.31) but it is unlikely that this represents a meaningful difference. Thus, it does not appear that changes in energy absorption strategies differed markedly in those with larger BMIs compared to those with smaller BMIs. However, it would be worth re-examining these patterns by equalizing task difficulty relative to sex differences in strength.

In summary, equalizing the relative task difficulty between BMI-matched males and females did not induce the expected alterations in energy absorption strategies. During a standardized task where sex differences in knee energy absorption are typically present, the males and females in the current study performed similarly. When the task difficulty was increased to account for sex differences in lower extremity lean mass relative to total body mass, the males responded by increasing total energy absorption without the preferential reliance on the knee that was expected. When females were also asked to land from the increased height, which represented an exaggerated increase in relative task difficulty compared to the males, the females responded to the greater height in a manner nearly identical to that of the males. These findings may suggest that the relative task difficulty was underestimated in this highly athletic and skilled population and that further work is needed to determine the most appropriate equalizing scheme for creating parity in task difficulty between males and females.

Also noted in this study was that total energy absorption from the biomechanical model only accounted for 56.2±10.5% of the total potential energy associated with the 0.45m drop height during the Height_{STD} condition in females and 68.5±9.8% in males. The amount of energy absorption for which the current biomechanical models did not account include factors such as energy absorption through the passive structures (e.g. bones, ligaments, etc.), through segments not modeled (i.e. the trunk and upper extremity), and in other planes of motion (i.e. frontal and transverse). Additionally, energy dissipation in the wobbling mass (i.e. fat and other non-contractile soft tissue) appears to be a significant factor when attempting to account for the total amount of the kinetic energy introduced during an impact; including an estimate of the energy dissipated in the wobbling mass can account for more energy than that accounted for in a rigid segment model alone (Pain & Challis, 2006). Energy dissipation by the soft tissues may be an especially relevant factor in the current context considering the greater fat mass in females compared to males, particularly in the lower extremity, which would be consistent with the 10% decrement in energy accounted for in females compared to males. Specifically, since females typically possess greater fat mass in the lower extremity, while males possess more fat mass in the trunk (Fuller, Laskey, & Elia, 1992; Malina, 2005), energy dissipated at these sites would have preferentially influenced the measurements in the females since we only modeled the lower extremity. Therefore, any energy lost in the trunk in males due to soft tissue vibration would not have even been a factor in the current calculations.

The magnitude of energy unaccounted for in this study is substantially larger than that found in a previous study which accounted for 79.1±17.5% and 78.1±16.4% of the total potential energy during the same task in females and males, respectively (Schmitz & Shultz, 2010). Small differences between studies may explain the inconsistency. For example, the subjects in the previous study were barefoot and it appears that energy absorption is slightly greater, albeit non-significant, when barefoot vs. shod (Shultz , et al., In Review). This small difference combined with other factors may collectively account for the discrepancy. One particular factor may be differences in instrumentation. Specifically that the motion capture system in the current study uses markers that may be more susceptible to accessory motion on the skin compared to the sensors used in the former study which were secured more closely to rigid bony landmarks. Movement artifact may result in an underestimation in energy absorption and appear as energy lost due to vibration of the markers.

Methodologically, additional considerations that have come to light are the normalization procedures used in this study. Consistent with the existing literature (Zhang, et al., 2000; Decker, et al., 2003; Kulas, et al., 2008; Zhang, et al., 2008; Norcross, et al., 2010; Schmitz & Shultz, 2010; Shultz, et al., 2010), energy absorption was modeled in a single limb during a double-leg landing, but was normalized to total body weight. Working on the assumption that energy absorption responsibilities were evenly distributed between both limbs, this may be resulting in an underestimation of the eccentric work being performed. Hence, it may be more appropriate to normalize to ½ body weight. Although this normalization procedure would just result in a linear transformation of the values, it would increase the functional interpretation of the measurement and should be considered in the future. Additionally, all variables used in the statistical analyses were normalized to total body mass. Predicting normalized energy absorption values with normalized lean mass and eccentric strength values may additionally underestimate the predictions by continually reducing the inter-subject differences and may have biased the predictions. Perhaps a better method would be to account for body mass as a predictor variable, instead of continually normalizing each individual variable. These limitations in normalization procedures deserve merit and consideration in future theoretical designs; addressing them may increase the interpretability of biomechanical outcomes as well as statistical analyses.

Chapter Summary

In this study, two methods were used to investigate the influence of strength on lower extremity energy absorption strategies. First, the extent to which strength predicted energy absorption in males and females was examined in relation to task difficulty. Then, based on differences in lower extremity lean mass, the experimental landing condition was modified to equalize the relative task difficulty between males and females. The overall finding was that relationships between eccentric strength and energy absorption were present in females, but not males. However, the extent to which eccentric strength and energy absorption were related was not dependent on the level of task difficulty. Additionally, it does not appear that sex differences in strength, thus relative task difficulty, are responsible for the previously-observed sex differences in energy absorption strategies.

Although strength appears to be a factor in energy absorption capabilities, strength does not become a more important determinant as the muscular demands of the task increase. It is unclear why stronger relationships between strength and energy absorption seem to exist in females, but not males. It appears that females prefer to perform landing maneuvers in a manner that relies more on muscle strength, whereas males rely on other factors such as coordination. More work is needed to elucidate other possible factors that have the potential to influence sex differences in the relationship between strength and energy absorption capabilities, and further, sex differences in energy absorption strategies. Identifying these factors may have significant implications for more focused prevention and training strategies and may point to a need for sex-specific interventions.

Additionally, because there was still a large portion of energy absorption which was not explained by lean mass or strength, there are still unidentified factors which influence the ability to safely decelerate the body during landing and account for sex differences in landing strategies. As the typical sex differences in energy absorption were not as evident in the current population studied, more work is needed to investigate the influence of experience and training on energy absorption strategy. Further work is needed to elucidate additional mechanisms by which energy absorption strategies may be modulated and how energy absorption strategies are related to injury risk across sex. This includes examining possible relationships between sagittal plane energy absorption and frontal/transverse plane knee joint loading (Pollard, et al., 2010), which is most often related to ACL injury. Additionally, as recent evidence suggests that the timing, not the just the magnitude, of energy absorption patterns may be related to biomechanical factors which have been linked to ACL injury (Norcross, et al., 2010), this also points to the need to investigate time-related patterns of muscle activation and energy absorption. Greater understanding of these factors will further inform our training and intervention strategies with the ultimate goal of reducing ACL injury risk.

As the effect of body composition on landing energetics was addressed indirectly in this study via the effect of lean mass on eccentric strength, further work is still needed to more directly investigate the influence of regional body composition and proportion on landing mechanics. As the largest sex differences in body composition are in the upper extremity (Malina, 2005), with males possessing a larger proportion of total body mass in the HAT (head, arms, trunk) segment (Fuller, et al., 1992), this sex difference likely has implications for the lower extremity. Specifically, the greater relative mass of the HAT segment in males results in a larger moment of inertia and may explain the greater hip flexion and extensor moments at the hip during landing compared to females as the males attempt to control the forward and downward motion of the heavier trunk. As greater trunk flexion is thought to reduce loading at the knee (Blackburn & Padua, 2008; Kulas, et al., 2008; Blackburn & Padua, 2009; Kulas, et al., 2010), this sexually dimorphic anthropometric characteristic may help explain the "hip-dominant" strategies observed in males. Further, these strategies may protect the knee from excessive loading (Powers, 2010) and potentially provide some explanation for the reduced incidence of ACL injury compared to females.

While the findings from this study have increased our knowledge of the relationships between body composition, strength, and energy absorption strategies, this work is not without limitations. These findings are limited to a healthy, highly athletic population and cannot be extended to other populations without further investigation. The validity of the measurements made in this study was dependent upon maximal effort of the subjects during strength and biomechanical testing. In particular, the validity of the eccentric strength testing protocol (which was difficult to perform) is completely dependent on the maximal effort of the subjects in order to accurately capture the true force-producing capabilities of the available musculature. Moreover, the validity of the peak torque measurements from the strength testing protocol are only specific to the particular knee flexion angle at which the peak torque measurement was captured. Additionally, the experimental task used in the current investigation was primarily sagittal in nature. Since ACL injury likely results from a combination of sagittal, frontal, and transverse plane loading, future work should investigate the influence of strength on controlling multiplanar motion. Finally, because the findings indicate that a range of strategies may be available to an athletic population when performing a complex task, including an analysis of muscle activation and coordination would provide additional insight.

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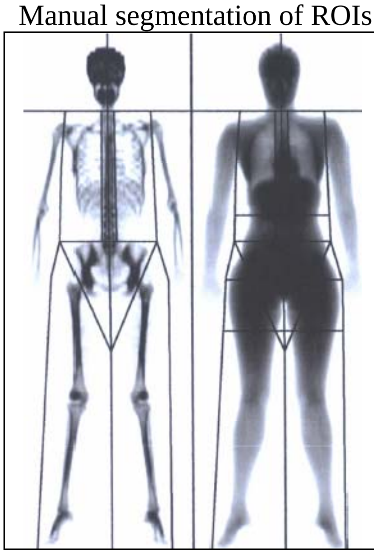
APPENDIX A: LOWER EXTREMITY LEAN MASS (LELM) MEASUREMENTS

Anatomical landmarks for regions of interest (ROI) placement

Regions	Proximal	Distal	Medial	Lateral
Arms	Soft tissue	End of longest finger	GH joint space	Soft tissue
Legs	Femoral neck	End of longest toe	Soft tissue	Soft tissue
Trunk	Border of soft tissue on chin	Soft tissue of pelvis	N/A	GH joint space to greater trochanter
Calf	Lateral joint line of the knee	Tip of the lateral malleolus	Soft tissue	Soft tissue
Foot	Tip of the lateral malleolus	End of longest toe	Soft tissue	Soft tissue
Thigh	Tip of greater trochanter	Distal border of femoral condyle @ lateral joint line	Soft tissue	Soft tissue

From Montgomery MM, Wideman L, Shultz SJ. Reliability of manual definitions of anatomical regions on total body DXA scans. Unpublished data.

Manual segmentation of ROIs and resulting Body Composition measurements



BODY COMPOSITION

Region	Tissue[1] (%Fat)	Region (%Fat)	Tissue[1] (g)	Fat[1] (g)	Lean[1] (g)	BMC (g)	Total Mass (kg)
Left Arm	17.0	16.0	2,278	386	1,892	132	2.4
Left Leg	32.6	31.2	9,817	3,201	6,616	449	10.3
Left Trunk	21.5	20.9	12,986	2,793	10,193	363	13.3
Left Total	24.9	23.9	26,319	6,557	19,761	1,112	27.4
Right Arm	16.9	16.0	2,348	398	1,950	143	2.5
Right Leg	32.6	31.2	10,277	3,355	6,922	467	10.7
Right Trunk	21.5	20.9	13,372	2,872	10,500	388	13.8
Right Total	24.5	23.4	28,626	7,001	21,625	1,321	29.9
Arms	17.0	16.0	4,626	784	3,841	275	4.9
Legs	32.6	31.2	20,094	6,556	13,538	917	21.0
Trunk	21.5	20.9	26,358	5,666	20,693	750	27.1
Android	21.0	20.6	3,393	711	2,681	55	3.4
Gynoid	38.6	37.7	9,920	3,830	6,090	242	10.2
Total	24.7	23.6	54.945	13.559	41.386	2.432	57.4

APPENDIX B: CALCULATION OF RELATIVE TASK DEMANDS

Drop Jump Landing Height calculations for Equalization of Relative Demands

	Female		Male
1.	$\frac{\text{LELM}}{\text{PE}}$	$=$	$\frac{\text{LELM}}{\text{PE}}$
2.	$\frac{\text{LELM}}{\text{PE}}$	$=$	$\frac{\text{LELM}}{\text{mgh}}$
3.	$\left(\frac{\text{LELM}}{\text{PE}}\right) * \text{h}$	$=$	$\frac{\text{LELM}}{\text{mg}}$
4.	Height	$=$	$\frac{\text{LELM}}{\left(\frac{\text{LELM}}{\text{PE}}\right)(\text{mg})}$

APPENDIX C: COUNTERBALANCING SCHEME FOR EACH MALE-FEMALE PAIR

PAIR ID#	Drop Jump Condition	
1	Standardized	Equalized
2	Equalized	Standardized
3	Standardized	Equalized
4	Equalized	Standardized
5	Standardized	Equalized
6	Equalized	Standardized
7	Standardized	Equalized
8	Equalized	Standardized
9	Standardized	Equalized
10	Equalized	Standardized
11	Standardized	Equalized
12	Equalized	Standardized
13	Standardized	Equalized
14	Equalized	Standardized
15	Standardized	Equalized
16	Equalized	Standardized
17	Standardized	Equalized
18	Equalized	Standardized
19	Standardized	Equalized
20	Equalized	Standardized
21	Standardized	Equalized
22	Equalized	Standardized
23	Standardized	Equalized
24	Equalized	Standardized
25	Standardized	Equalized
26	Equalized	Standardized
27	Standardized	Equalized
28	Equalized	Standardized
29	Standardized	Equalized
30	Equalized	Standardized
31	Standardized	Equalized
32	Equalized	Standardized
33	Standardized	Equalized
34	Equalized	Standardized
35	Standardized	Equalized
36	Equalized	Standardized

APPENDIX D: STANDARD DYNAMIC FLEXIBILITY WARM-UP

Description of Dynamic Flexibility Exercises*

Heel-Toe Walk
Step forward into a single-leg balance position. Plantar flex ankle to perform calf raise while flexing opposite hip and pulling knee towards chest drawing the knee towards the chest, with the foot dorsi-flexed. Step forward and repeat with opposite leg

Walking Calf Stretch
Begin the activity by stepping forward an exaggerated stride length and moving the upper torso forward over the front foot. Press the heel of the rear foot into the ground and extend the knee fully to maximally stretch the gastrocnemius muscle of the rear leg then release and step forward/backwards into alternate stride.

Alternate Leg Heel Kicks
While keeping knees pointing towards the ground, kick heel up towards the glutes (easy). Switch feet and repeat while moving forwards. To progress (progessive), repeat with a more brisk kick, touching glutes with heels.

Walking Quadriceps Stretch
Step forward into a single-leg balance stance while drawing the opposite heel upwards. Use ipsilateral hand to grasp foot and extend and raise the leg behind the body while simultaneously leaning forward with the trunk to maximally stretch the muscles of the quadriceps and hip flexors. On reaching maximal stretch, release leg and step forward to repeat with alternate foot.

Walking Hamstrings
Step forwards or backwards a distance equivalent to approximately 1/2 a stride length and extend the knee fully while dorsi-flexing the foot maximally. Crossing the arms to ensure that hips do not rotate bend at the waist until hamstring muscles are stretched maximally. On reaching maximal stretch stand upright and step forward with the opposite foot.

Walking Toe Touch (for hamstrings)
Begin by stepping forward a normal stride length to place the foot flat on the ground. Extend the knee fully and begin to bend at the waist. With both hands in front of the body continue to bend at the waist while simultaneously lifting the opposite leg behind the body and extending the hip. Continue to swing the opposite leg up and behind the body while maintaining balance and touching the ground until a maximal stretch is achieved in the hamstring of the support leg. Return the leg to the ground and step forward with the opposite leg to repeat.

Single Leg Kicks
Step forward into a single-leg stance. With the knee slightly flexed and the foot dorsi-flexed, kick the opposite leg directly upwards towards the chest, using maximal hip flexion range of motion. Coordinate the swing of the opposite arm to touch the toes of the swing leg with the opposite hand. Lower the leg to the ground and step forward to repeat with the alternate leg.

Walking Side Lunge (alternating legs)
Begin movement by stepping laterally into a side-lunge position. Drop down and backwards at the hips flexing at the front knee and hip to drop as deep as possible. Maintain extension of the rear leg, keeping the foot on the ground. Move as deep as possible and thus into a maximal stretch of the adductors. Arms move forward in coordination with the movement down and back through the hips of the lead leg. Upon reaching maximal stretch drive upwards via extension of the hip-knee-ankle of the lead foot to turn and stretch the alternate leg in the same fashion.

Open the Gate
Begin movement by stepping forwards into a single-leg balance position. Draw the opposite leg towards the mid-line of the body and upwards. Maintaining a flexed knee bring the leg maximally upwards, then swing the leg wide via abduction to maximally stretch the adductors. The knee remains flexed throughout the movement, the foot lightly dorsi-flexed.

Close the Gate
Begin movement by stepping forward across the body and onto a single foot. Drive the opposite leg upwards, the knee slightly flexed and the foot dorsi-flexed. Continue movement of the leg upwards and across the body, with the upper torso turning in the opposite direction. Coordinate movement of the arms and upper torso against the movement of the stretched leg across the body in order to reach a maximal stretch. Lower the stretched leg to the ground and across the body to crossover the foot of the stance leg. Step forward to stretch the alternate leg.

Leg Swings
Place one hand on a wall or a partner for support, perform full range of motion leg swings (hip flexion to hip extension) to alternately stretch hip and knee flexors and extensors.

* Adapted with permission from Cone Fitness Training and Consulting, LLC.

APPENDIX E: STRENGTH TESTING RESULTS

E1. Peak torque values normalized to total body mass. Mean±SD and range (min-max) are provided for all values normalized to total body mass (Nm/kg).

	ALL SUBJECTS (n=70)	FEMALES (n=35)	MALES (n=35)	p-value	Effect Size d
Quad MVIC	3.1±0.6 (2.1-4.5)	2.8±0.4 (2.1-3.6)	3.4±0.6* (2.4-4.5)	p<.001	1.00
Quad 60 Con#	2.9±0.6 (1.7-4.7)	2.6±0.4 (1.7-3.5)	3.4±0.5* (2.6-4.7)	p<.001	1.52
Quad 60 Ecc#	3.6±0.7 (2.5-5.5)	3.3±0.5 (2.5-4.7)	4.0±0.7* (2.7-5.5)	p<.001	1.01
Quad 180 Con	2.6±0.6 (1.5-4.0)	2.2±0.4 (1.5-3.2)	3.0±0.4* (2.2-4.0)	p<.001	1.81
Quad 180 Ecc	3.5±0.7 (1.8-5.7)	3.2±0.5 (1.8-4.0)	3.9±0.7* (2.0-5.7)	p<.001	0.99
Ham MVIC	1.5±0.4 (0.7-2.4)	1.3±0.3 (0.7-1.8)	1.8±0.3* (1.1-2.4)	p<.001	1.70
Ham 60 Ecc	2.3±0.4 (1.5-3.3)	2.0±0.3 (1.5-2.6)	2.6±0.3* (2.1-3.3)	p<.001	1.82
Ham 60 Con	1.7±0.4 (1.0-2.4)	1.4±0.2 (1.0-1.9)	1.9±0.3* (1.4-2.4)	p<.001	1.89
Ham180 Ecc	2.2±0.4 (1.3-3.1)	2.0±0.3 (1.3-2.5)	2.5±0.3* (2.0-3.1)	p<.001	2.07
Ham 180 Con	1.5±0.4 (0.7-2.2)	1.2±0.3 (0.7-1.7)	1.8±0.2* (1.3-2.2)	p<.001	2.28

* Indicates Males > Females (p<.05)

Males (n=27) for this measurement

E2. Peak torque values normalized to lower extremity lean mass. Mean±SD and Range (min-max) are provided for all values normalized to lower extremity lean mass (Nm/kg).

	ALL SUBJECTS (n=70)	**FEMALES (n=35)**	**MALES (n=35)**	**p-value**	**Effect Size d**
Quad MVIC	24.0±3.4 (16.3-32.8)	24.5±3.4 (17.9-32.8)	23.5±3.3 (16.3-29.0)	p=.216	0.30
Quad 60 Con #	22.9±3.4 (17.4-32.8)	22.2±2.8 (18.1-30.0)	23.9±3.8* (17.4-32.8)	p=.048	0.45
Quad 60 Ecc #	28.5±4.5 (19.6-40.1)	28.6±4.0 (22.1-37.6)	28.4±5.1 (19.6-40.1)	p=.845	0.04
Quad 180 Con	19.7±3.0 (14.8-26.8)	18.9±3.0) (14.8-26.8)	20.5±2.8* (15.0-26.8)	p=.023	0.57
Quad 180 Ecc	27.1±4.8 (14.3-40.4)	27.2±4.7 (18.2-40.4)	27.0±4.9 (14.3-35.4)	p=.824	0.05
Ham MVIC	11.8±2.0 (6.8-16.0)	11.2±1.7 (6.8-13.8)	12.4±2.0* (7.3-16.0)	p=.009	0.60
Ham 60 Ecc	17.7±2.1 (12.8-22.2)	17.5±1.9 (12.8-21.0)	17.9±2.3 (13.6-22.2)	p=.361	0.20
Ham 60 Con	12.7±1.9 (8.4-16.5)	12.1±1.6 (8.4-15.2)	13.3±2.0* (9.1-16.5)	p=.006	0.62
Ham180 Ecc	16.9±2.1 (11.1-21.6)	16.5±2.1 (11.1-20.6)	17.2±2.1 (12.4-21.6)	p=.187	0.32
Ham 180 Con	11.5±2.0 (6.2-15.0)	10.7±1.9 (6.2-14.4)	12.4±1.8* (8.1-15.0)	p<.001	0.98

* Indicates Males > Females (p<0.05)
# Males (n=27) for this measurement

E3. Day-to-day reliability for peak torque values on familiarization day vs. testing day (n=15).

PEAK TORQUE (Nm)						
	Day 1		**Day 2**		**$ICC_{2,k}$**	**SEM**
	Mean	**SD**	**Mean**	**SD**		
QuadMVIC	206.36	36.75	204.77	37.22	0.93	9.80
Quad60con	207.43	52.54	210.05	54.92	0.98	7.71
Quad60ecc	256.83	52.74	263.63	64.64	0.94	14.07
Quad180con	170.46	50.62	180.18	50.38	0.97	8.54
Quad180ecc	239.73	56.08	241.27	47.97	0.91	15.99
HamMVIC	111.73	30.47	105.92	29.20	0.96	6.02
Ham60ecc	158.86	40.09	156.63	36.90	0.98	5.15
Ham60con	111.39	35.06	112.40	30.88	0.97	5.55
Ham180ecc	147.76	41.76	149.55	36.77	0.96	7.79
Ham180con	98.60	37.53	102.89	32.88	0.96	6.69
NORMALIZED PEAK TORQUE (Nm/kg)						
	Day 1		**Day 2**		**$ICC_{2,k}$**	**SEM**
	Mean	**SD**	**Mean**	**SD**		
QuadMVIC	2.99	0.47	2.96	0.35	0.86	0.16
Quad60con	2.97	0.53	3.01	0.54	0.95	0.12
Quad60ecc	3.69	0.49	3.78	0.59	0.83	0.22
Quad180con	2.43	0.54	2.58	0.54	0.94	0.13
Quad180ecc	3.43	0.54	3.48	0.43	0.72	0.26
HamMVIC	1.60	0.36	1.53	0.34	0.94	0.08
Ham60ecc	2.28	0.43	2.26	0.38	0.97	0.07
Ham60con	1.59	0.38	1.62	0.33	0.95	0.08
Ham180ecc	2.11	0.44	2.15	0.37	0.90	0.13
Ham180con	1.40	0.43	1.47	0.36	0.93	0.11

APPENDIX F: PEARSON PRODUCT CORRELATIONS FOR ALL STRENGTH AND ENERGY ABSORPTION VALUES

F1 **ALL SUBJECTS**

	$HEIGHT_{STD}$				$HEIGHT_{EQU}$			
	Total	**Hip**	**Knee**	**Ankle**	**Total**	**Hip**	**Knee**	**Ankle**
QuadMVIC	0.413*	**0.327***	0.248*	0.022	0.423*	**0.304***	0.219	0.146
Quad60con	0.427*	0.331*	0.210	0.121	0.507*	0.133	0.430*	0.158
Quad60ecc	**0.555***	0.121	**0.470***	0.176	0.538*	0.144	**0.470***	0.135
Quad180con	0.547*	0.145	0.440*	0.189	**0.567***	0.213	0.439*	0.144
Quad180ecc	0.553*	0.199	0.425*	0.123	0.444*	0.271*	0.310*	0.054
HamMVIC	0.334*	0.450*	0.048	0.090	0.320*	**0.464***	0.003	0.159
Ham60ecc	0.433*	0.406*	0.132	0.193	**0.401***	0.384*	0.095	**0.253***
Ham60con	**0.441***	0.439*	0.106	0.220	0.395*	0.431*	0.052	0.270*
Ham180ecc	0.405*	**0.468***	0.123	0.078	0.349*	0.445*	0.065	0.121
Ham180con	0.393*	0.465*	0.099	0.105	0.356*	0.470*	0.049	0.141

* Significant correlation (p<0.05)

Bold denotes strongest significant correlation with EA variable.

F2 **FEMALES**

	$HEIGHT_{STD}$				$HEIGHT_{EQU}$			
	Total	**Hip**	**Knee**	**Ankle**	**Total**	**Hip**	**Knee**	**Ankle**
QuadMVIC	0.257	-0.087	0.267	0.087	0.163	-0.107	0.209	0.053
Quad60con	0.446	-0.077	0.543*	-0.075	0.332	-0.081	0.505*	-0.145
Quad60ecc	0.253	-0.156	0.352*	-0.021	0.132	-0.189	0.310	-0.097
Quad180con	**0.528***	-0.025	**0.583***	-0.039	**0.434***	-0.010	**0.545***	-0.086
Quad180ecc	0.426*	0.008	0.389*	0.088	0.336*	0.033	0.351*	0.010
HamMVIC	0.310	0.393*	0.194	-0.118	0.309	0.359*	0.209	-0.098
Ham60ecc	0.389*	0.282	0.235	0.057	0.349*	0.221	0.236	0.060
Ham60con	**0.490***	0.382*	0.296	0.046	**0.464***	0.344*	0.314	0.031
Ham180ecc	0.310	0.478*	0.182	-0.174	0.274	0.435*	0.163	-0.163
Ham180con	0.417*	**0.557***	0.242	-0.154	0.378*	**0.512***	0.243	-0.174

* Significant correlation (p<0.05)

Bold denotes strongest significant correlation with EA variable.

F3	MALES							
	HEIGHT$_{STD}$				HEIGHT$_{EQU}$			
	Total	**Hip**	**Knee**	**Ankle**	**Total**	**Hip**	**Knee**	**Ankle**
QuadMVIC	0.311	**0.336***	0.111	-0.032	0.365*	0.296	0.178	0.039
Quad60con	0.357	-0.215	0.427*	0.138	0.389*	-0.177	0.409*	0.180
Quad60ecc	**0.564***	0.028	**0.484***	0.171	**0.631***	0.059	**0.569***	0.125
Quad180con	0.306	-0.104	0.374*	-0.006	0.453*	-0.093	0.470*	0.074
Quad180ecc	0.193	0.132	0.203	-0.208	0.296	0.150	0.263	-0.124
HamMVIC	-0.104	0.241	-0.254	-0.006	-0.098	0.286	-0.315	0.132
Ham60ecc	0.090	0.197	-0.076	0.089	0.069	0.172	-0.126	0.228
Ham60con	0.022	0.212	-0.191	0.169	-0.040	0.200	-0.277	0.292
Ham180ecc	0.045	0.205	-0.080	0.000	-0.053	0.157	-0.178	-0.084
Ham180con	-0.138	0.127	-0.226	0.030	-0.159	0.161	-0.304	0.131

* Significant correlation ($p<0.05$)

Bold denotes strongest significant correlation with EA variable.

APPENDIX G: INDIVIDUAL PATH RESULTS FROM MEDIATION ANALYSIS

G1. Results for mediation analysis, predicting energy absorption with lean mass, after controlling for eccentric quadriceps torque (All Subjects: n=70)

Path	Unstandardized Beta	Standard Error	Standardized Beta	t-value	p-value
Total Energy Absorption ($J*N^{-1}*m^{1}$)					
C	0.317	0.085	0.413	3.743	0.000*
A	11.754	1.924	0.595	6.110	0.000*
B	0.011	0.005	0.284	2.120	0.038*
C'	0.187	0.103	0.244	1.822	0.073
Hip Energy Absorption ($J*N^{-1}*m^{-1}$)					
C	0.139	0.049	0.327	2.849	0.006*
A	11.754	1.924	0.595	6.110	0.000*
B	0.002	0.003	0.089	0.619	0.538
C'	0.117	0.061	0.274	1.910	0.060
Knee Energy Absorption ($J*N^{-1}*m^{-1}$)					
C	0.172	0.081	0.248	2.112	0.038*
A	11.754	1.924	0.595	6.110	0.000*
B	0.008	0.005	0.233	1.616	0.111
C'	0.076	0.100	0.109	0.755	0.453
Ankle Energy Absorption ($J*N^{-1}*m^{-1}$)					
C	0.008	0.044	0.022	0.181	0.857
A	11.754	1.924	0.595	6.110	0.000*
B	0.001	0.003	0.043	0.282	0.779
C'	-0.001	0.055	-0.004	-0.023	0.981

* Significant path coefficient ($p<0.05$)

G2. Results for mediation analysis, predicting energy absorption with lean mass, after controlling for eccentric quadriceps torque (Females: n=35)

Path	Unstandardized Beta	Standard Error	Standardized Beta	t-value	p-value
Total Energy Absorption ($J*N^{-1}*m^{1}$)					
C	0.294	0.156	0.312	1.884	0.068
A	9.662	2.826	0.511	3.419	0.002*
B	0.018	0.009	0.361	1.953	0.060
C'	0.120	0.174	0.127	0.688	0.497
Hip Energy Absorption ($J*N^{-1}*m^{-1}$)					
C	-0.023	0.078	-0.052	-0.298	0.768
A	9.662	2.826	0.511	3.419	0.002*
B	0.001	0.005	0.756	0.228	0.821
C'	-0.034	0.093	-0.485	-0.369	0.715
Knee Energy Absorption ($J*N^{-1}*m^{-1}$)					
C	0.364	0.144	0.402	2.520	0.017*
A	9.662	2.826	0.511	3.419	0.002*
B	0.012	0.009	0.249	1.360	0.183
C'	0.249	0.166	0.274	1.498	0.144
Ankle Energy Absorption ($J*N^{-1}*m^{-1}$)					
C	-0.043	0.087	-0.087	-0.500	0.621
A	9.662	2.826	0.511	3.419	0.002*
B	0.005	0.005	0.179	0.882	0.385
C'	-0.089	0.101	-0.178	-0.879	0.386

* Significant path coefficient ($p<0.05$)

G3. Results for mediation analysis, predicting energy absorption with lean mass, after controlling for eccentric quadriceps torque (Males: n=35)

Path	Unstandardized Beta	Standard Error	Standardized Beta	t-value	p-value
Total Energy Absorption ($J*N^{-1}*m^{1}$)					
C	0.046	0.197	0.041	0.234	0.816
A	9.846	5.294	0.308	1.860	0.072
B	0.007	0.007	0.199	1.092	0.283
C'	-0.023	0.206	-0.021	-0.113	0.911
Hip Energy Absorption ($J*N^{-1}*m^{-1}$)					
C	0.179	0.124	0.244	1.444	0.158
A	9.846	5.294	0.308	1.860	0.072
B	0.001	0.004	0.063	0.348	0.730
C'	0.165	0.132	0.225	1.248	0.221
Knee Energy Absorption ($J*N^{-1}*m^{-1}$)					
C	0.048	0.197	0.042	0.242	0.810
A	9.846	5.294	0.308	1.860	0.072
B	0.008	0.007	0.210	1.157	0.256
C'	-0.026	0.206	-0.023	-0.125	0.901
Ankle Energy Absorption ($J*N^{-1}*m^{-1}$)					
C	-0.177	0.087	-0.332	-2.025	0.051
A	9.846	5.294	0.308	1.860	0.072
B	-0.002	0.003	-0.117	-0.672	0.507
C'	-0.158	0.093	-0.296	-1.703	0.098

* Significant path coefficient ($p<0.05$)

G4. Results for mediation analysis, predicting energy absorption with lean mass, after controlling for eccentric hamstring torque (All Subjects: n=70)

Path	Unstandardized Beta	Standard Error	Standardized Beta	t-value	p-value
Total Energy Absorption ($J*N^{-1}*m^{1}$)					
C	0.317	0.085	0.413	3.743	<0.001*
A	7.229	0.919	0.690	7.869	<0.001*
B	0.017	0.011	0.229	1.517	0.134
C'	0.195	0.116	0.255	1.687	0.096
Hip Energy Absorption ($J*N^{-1}*m^{-1}$)					
C	0.139	0.049	0.327	2.849	0.006*
A	7.229	0.919	0.690	7.869	<0.001*
B	0.019	0.006	0.464	3.112	0.003*
C'	0.003	0.064	0.006	0.041	0.968
Knee Energy Absorption ($J*N^{-1}*m^{-1}$)					
C	0.172	0.081	0.248	2.112	0.038*
A	7.229	0.919	0.690	7.869	<0.001*
B	-0.006	0.011	-0.092	-0.563	0.576
C'	0.216	0.113	0.312	1.909	0.061
Ankle Energy Absorption ($J*N^{-1}*m^{-1}$)					
C	0.008	0.044	0.022	0.181	0.857
A	7.229	0.919	0.690	7.869	<0.001*
B	0.004	0.006	0.121	0.717	0.476
C'	-0.022	0.060	-0.061	-0.365	0.717

* Significant path coefficient (p<0.05)

G5. Results for mediation analysis, predicting energy absorption with lean mass, after controlling for eccentric hamstring torque (Females: n=35)

Path	Unstandardized Beta	Standard Error	Standardized Beta	t-value	p-value
Total Energy Absorption ($J*N^{-1}*m^{1}$)					
C	0.294	0.156	0.312	1.884	0.068
A	5.856	1.460	0.573	4.012	<0.001*
B	0.018	0.019	0.196	0.971	0.339
C'	0.188	0.190	0.199	0.987	0.331
Hip Energy Absorption ($J*N^{-1}*m^{-1}$)					
C	-0.023	0.078	-0.052	-0.298	0.768
A	5.856	1.460	0.573	4.012	<0.001*
B	0.033	0.007	0.756	4.477	<0.001*
C'	-0.219	0.076	-0.485	-2.871	0.007*
Knee Energy Absorption ($J*N^{-1}*m^{-1}$)					
C	0.364	0.144	0.402	2.520	0.017*
A	5.856	1.460	0.573	4.012	<0.001*
B	-0.006	0.017	-0.072	-0.364	0.718
C'	0.401	0.179	0.443	2.247	0.032*
Ankle Energy Absorption ($J*N^{-1}*m^{-1}$)					
C	-0.043	0.087	-0.087	-0.500	0.621
A	5.856	1.460	0.573	4.012	<0.001*
B	-0.009	0.010	-0.185	-0.870	0.391
C'	0.010	0.106	0.019	0.090	0.929

* Significant path coefficient ($p<0.05$)

G6. Results for mediation analysis, predicting energy absorption with lean mass, after controlling for eccentric hamstring torque (Males: n=35)

Path	Unstandardized Beta	Standard Error	Standardized Beta	t-value	p-value
Total Energy Absorption ($J*N^{-1}*m^{1}$)					
C	0.046	0.197	0.041	0.234	0.816
A	0.887	1.999	0.077	0.444	0.660
B	0.004	0.017	0.042	0.239	0.812
C'	0.042	0.200	0.037	0.212	0.834
Hip Energy Absorption ($J*N^{-1}*m^{-1}$)					
C	0.179	0.124	0.244	1.444	0.158
A	0.887	1.999	0.077	0.444	0.660
B	0.012	0.011	0.188	1.113	0.274
C'	0.168	0.124	0.229	1.359	0.184
Knee Energy Absorption ($J*N^{-1}*m^{-1}$)					
C	0.048	0.197	0.042	0.242	0.810
A	0.887	1.999	0.077	0.444	0.660
B	-0.008	0.017	-0.084	-0.477	0.637
C'	0.055	0.200	0.049	0.275	0.785
Ankle Energy Absorption ($J*N^{-1}*m^{-1}$)					
C	-0.177	0.087	-0.332	-2.025	0.051
A	0.887	1.999	0.077	0.444	0.660
B	0.001	0.008	0.026	0.153	0.879
C'	-0.178	0.089	-0.334	-2.001	0.054

* Significant coefficient ($p<0.05$)

G7. Full regression models for Hypothesis 1. Unstandardized coefficients and adjusted R^2 value are provided.

Dependent Variable	Predictor Variables		R^2 value	Final Regression Equation
Total Energy Absorption	LELM, Quad	All	0.200*	EA_{TOT}= .074 + 0.187(LELM) + 0.011(Quad) †
		Females	0.143*	EA_{TOT}= 0.065 + 0.120(LELM) + 0.018(Quad)
		Males	-0.029	EA_{TOT}= 0.153 -0.023(LELM)+0.007(Quad)
Hip Energy Absorption	LELM, Quad	All	0.085*	EA_{HIP}= -0.005 +0.117(LELM) + 0.002(Quad)
		Females	-0.058	EA_{HIP}= 0.032 - 0.034(LELM) + 0.001(Quad)
		Males	0.004	EA_{HIP}= -0.015 + 0.165(LELM) + 0.001(Quad)
Knee Energy Absorption	LELM, Quad	All	0.070*	EA_{KNEE}= 0.046 + 0.076(LELM) + 0.008(Quad)
		Females	0.158*	EA_{KNEE}= -0.005 + 0.248(LELM) + 0.012(Quad)
		Males	-0.018	EA_{KNEE}= 0.076 - 0.026(LELM) + 0.007(Quad)
Ankle Energy Absorption	LELM, Quad	All	-0.028	EA_{ANK} = 0.032 - 0.001(LELM) + 0.001(Quad)
		Females	-0.030	EA_{ANK} = 0.038 - 0.089(LELM) + 0.005(Quad)
		Males	0.068	EA_{ANK}= 0.090 - 0.158(LELM) - 0.002Quad)

Dependent Variable	Predictor Variables		R^2 value	Final Regression Equation
Total Energy Absorption	LELM, Ham	All	0.174*	EA_{TOT}= 0.073 + 0.195(LELM) + 0.017(Ham)
		Females	0.068	EA_{TOT}= 0.071 + 0.188(LELM) + 0.018(Ham)
		Males	-0.059	EA_{TOT}= 0.151 + 0.042(LELM) + 0.004(Ham)
Hip Energy Absorption	LELM, Ham	All	0.196*	EA_{HIP}= -0.009 + 0.003(LELM) + 0.019(Ham) †
		Females	0.349*	EA_{HIP}= 0.015 - 0.219(LELM)* + 0.168(Ham) †
		Males	0.038	EA_{HIP}= -0.040 + 0.168(LELM) + 0.012(Ham)
Knee Energy Absorption	LELM, Ham	All	0.038	EA_{KNEE}= 0.051 + 0.216(LELM) - 0.006(Ham)
		Females	0.113*	EA_{KNEE}= 0.009 + 0.401(LELM) † - 0.006(Ham)
		Males	-0.053	EA_{KNEE}= 0.102 + 0.055(LELM) - 0.008(Ham)
Ankle Energy Absorption	LELM, Ham	All	-0.022	EA_{ANK}= 0.031 - 0.02(LELM) + 0.004(Ham)
		Females	-0.030	EA_{ANK}= 0.047 + 0.010(LELM) - 0.009(Ham)
		Males	0.056	EA_{ANK}= 0.086 - 0.178(LELM) + 0.001(Ham)

* Indicates significant model ($p<0.05$)

† Indicates significant regression coefficient ($p<0.05$)

APPENDIX H: REGRESSION COEFFICIENTS FOR FINAL MODELS PREDICTING ENERGY ABSORPTION WITH SEX, STRENGTH, AND SEX*STRENGTH INTERACTIONS

H1. Regression coefficients from full model when predicting total energy absorption (EA_{TOT}; $J*N^{-1}*kg^{-1}$) with eccentric strength ($Quad_{ECC}$ and Ham_{ECC}; Nm/kg) during the $Height_{STD}$ and $Height_{EQU}$ conditions.

Variable	Unstandardized Beta	SE	Standardized Beta	t-value	p-value
$Height_{STD}$					
Intercept	0.033	0.031		1.037	0.304
$Quad_{ECC}$	0.021	0.008	0.539*	2.629	0.011
Ham_{ECC}	0.028	0.014	0.380*	2.020	0.048
Sex*$Quad_{ECC}$	-0.013	0.010	-0.258	-1.322	0.191
Sex*Ham_{ECC}	-0.023	0.023	-0.185	-1.006	0.318
$Height_{EQU}$					
Intercept	0.074	0.035		2.093	0.040
Sex	0.013	0.010	0.208	1.339	0.185
$Quad_{ECC}$	0.015	0.006	0.345*	2.709	0.009
Ham_{ECC}	0.022	0.017	0.263	1.278	0.206
Sex*Ham_{ECC}	-0.040	0.025	-0.293	-1.626	0.109

* Significant regression coefficient ($p<0.05$)

H2. Regression coefficients from full model when predicting hip energy absorption (EA_{HIP}; $J*N^{-1}*kg^{-1}$) with eccentric strength ($Quad_{ECC}$ and Ham_{ECC}; Nm/kg) during the $Height_{STD}$ and $Height_{EQU}$ conditions.

Variable	Unstandardized Beta	SE	Standardized Beta	t-value	p-value
$Height_{STD}$					
Intercept	-0.009	0.010		-0.926	0.358
Ham_{ECC}	0.019	0.004	0.468*	4.372	<0.001
$Height_{EQU}$					
Intercept	-0.006	0.011		-0.881	0.580
Ham_{ECC}	0.019	0.005	0.445*	2.961	<0.001

* Significant regression coefficient ($p<0.05$)

H3. Regression coefficients from full model when predicting knee energy absorption (EA_{KNEE}; J*N^{-1}*kg^{-1}) with eccentric strength ($Quad_{ECC}$ and Ham_{ECC}; Nm/kg) during the $Height_{STD}$ and $Height_{EQU}$ conditions.

Variable	Unstandardized Beta	SE	Standardized Beta	t-value	p-value
$Height_{STD}$					
Intercept	0.032	0.025		1.316	0.193
$Quad_{ECC}$	0.019	0.007	0.535*	2.526	0.014
Sex*$Quad_{ECC}$	-0.010	0.010	-0.208	-1.005	0.318
Sex*Ham_{ECC}	-0.015	0.015	-0.136	-1.037	0.304
$Height_{EQU}$					
Intercept	0.049	0.017		2.919	0.005
$Quad_{ECC}$	0.016	0.005	0.424*	3.312	0.001
Sex*Ham_{ECC}	-0.030	0.016	-0.245	-1.910	0.060

* Significant regression coefficient ($p<0.05$)

H4. Regression coefficients from full model when predicting ankle energy absorption (EA_{ANK}; J*N^{-1}*kg^{-1}) with eccentric strength ($Quad_{ECC}$ and Ham_{ECC}; Nm/kg) during the $Height_{STD}$ and $Height_{EQU}$ conditions.

Variable	Unstandardized Beta	SE	Standardized Beta	t-value	p-value
$Height_{STD}$					
Intercept	0.031	0.002		13.968	<0.001
Sex	0.007	0.003	0.252	2.000	0.050
Sex*$Quad_{ECC}$	-0.003	0.003	-0.142	-1.130	0.263
$Height_{EQU}$					
Intercept	0.032	0.002		13.295	<0.001
Sex	0.006	0.003	0.217	1.835	0.071